UNIVERSITAT POLITÈCNICA DE CATALUNYA

DEPARTAMENT ENGINYERIA ELECTRÒNICA

Remotely accessible demonstrator for dynamic characterization of pressure measurement catheters

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1. Introduction and purposes

1.1 Introduction

In this project an experimental board for the characterization of a catheter for blood pressure measurement was designed and implemented. This system aims to simulate an invasive blood pressure measurement system studying the static and dynamic behavior of this process. This project was thought and made in the laboratory of Instrumentation and Bioengineering of the Electronics Engineering Department in the UPC of Barcelona.

We wanted to study the time and frequency response of a pressure's measurement system using catheters. The idea is to see the static and dynamic comportment of this system introducing catheters with air, liquid or liquid with bubbles to find out the response. The static behavior was simulated with a stepper linear motor to verify the linearity of the system, instead the dynamic behavior was simulated with an inductive linear motor that produces an impulsive stimulus like the heartbeat.
The joint heartbeat-catheter system produces a damped second order response, of which is possible to study the damping factor and the oscillation frequency. This response can be modeled with an R-L-C circuit and the idea is to create a multiple catheter-sensor's system to find out the frequency response and trying to study the different returns with air, water and water with bubbles, to extract different behaviors with the damping introduced by the air.

At the end of each catheter we have put a piezoresistive sensor to obtain the transitory response, conditioned by an instrumentation amplifier for each sensor.
Measurements are made with an experimental board that permits to derive measures with remote connection and to drive motors using a clock signal got from the board and a driving software. This board is used to permit to remotely connect an user at a defined server for verifying measurement results from his computer for didactical purposes.

The following figure shows in general the block diagram of the remote laboratory:

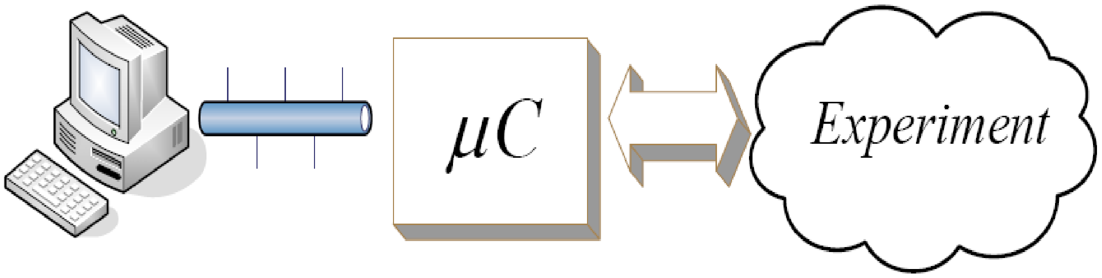


Figure 1: Block diagram of the remote laboratory

where the block Experiment is referred to the pressure measurement system.

1.2 Purposes

The purposes of this master thesis are the:

- Project and construction of a multiple measurement system for characterization of catheters
- Evaluation of the static and dynamic behavior using linear and inductive motors
- Provide a didactical instrumentation using the remote board for measurements
- Evaluate the transitory response, the frequency response and the R-L-C model to see the differences for the different situations with air, water and water with bubbles of air

1.3 Reasons

The principal reason of this project derive from the need to develop an experimental system, composed of motor system, circuit board and remote unit, to simulate the heartbeat and, consequently to make online pressure measurements using the remote board, providing a didactical tool for laboratory courses.

This derives from the study of invasive blood pressure measurements with catheter, used in medical ambit, evaluating the responses using a laboratory tool, without the use of approved medical components and allowing the visualization of real catheter-like signal in the classroom, without moving complex and heavy instrumentation.

1.4 Methods: tasks and tools

1.4.1 Tasks

These are the principal tasks followed for this project:

- Study and analysis of the invasive blood pressure measurement system, through manuals that describes the methods, dynamic behavior and sensor conditioning methods.
- Adaption and utilization of an existing circuit board to start the comprehension of the electrical problem. This comprehends the welding of components to adapt the circuit to our needs, learning of the first static response to verify the linearity of the system, also using a Labview software to make measurements.
- Use of the inductive motor, through a driving circuit, to start to see the first dynamic response with air in the catheters.
- Insertion of water in the catheters to study dynamic behavior, first hitting a syringe placed at the end of the catheters, and then using the inductive motor.
- Design of the final circuit and development of the circuit board, first choosing electronic components (sensors, amplifiers, resistances, capacitors..), then projecting with computer programs.
- Measurements to verify the correct behavior of the system
- Signal processing to extract the frequency response

1.4.2 Tools

These are the tools used for this project:

- Power supply sources, oscilloscope, multimeter and other laboratory tools
- Orcad Capture Cis 9.2 to project the schematic of material practice
- Ultiboard 10.1 to project the layout of the circuit board
- Labview 8.2 to program the software for driving the motors and measuring the circuit responses

1.5 Chapters' content

In the second chapter is described the theory of the arterial pressure measurement, first talking about the history of this measurements and portraying at physiologic level. Then there will be a description of invasive blood pressure measurements, and finally a look to the dynamic characteristic of catheter-sensor system

In third chapter there is an analysis of the problem, describing the adjustments made in the preliminary set-up, through the description of changes to adapt the system with laboratory tools and describing the static and dynamic characterization of the system. Then there will be a block diagram that describes the process.

In the fourth chapter we'll talk of the design of the circuit and the board, describing the remote board (hardware, software, and driver Labview), the system components (sensors, amplifiers, motors, and drivers) and first testing board with some photos and results.

In the fifth chapter there will be a description of the realization of the final board. In the first part of this chapter we'll talk about the schematic and the layout also introducing some photos. Then there will be a description of the control software for static and dynamic measurements

In the sixth chapter we'll describe the measurements with air, water and water with bubbles to verify the damping of system in several conditions.

Finally in the seventh chapter there will be the conclusions of the project and the future perspectives of the developed system.

2 Arterial pressure measurements

2.1 History and physiological description

The measurement of physiological fluid pressures is of interest to both biomedical researchers and medical clinicians. Obviously with fluid we'll be able to refer to liquids and gases, but here we'll talk only about liquids. The most common pressure measurement is arterial blood pressure, which is almost routinely monitored by electronic instruments.

Blood pressure values in the various chambers of the heart and in the peripheral vascular system help the physician determine the functional integrity of the cardiovascular system. A number of direct (invasive) and indirect (non invasive) techniques are being used to measure blood pressure in the human. The function of blood circulation is to transport oxygen and other nutrients to the tissues of the body and to carry metabolic waste products away from the cells.

The heart is divided into two pumping systems, the right and the left side of the heart. These two pumps and their associated valves are connected to the pulmonary circulation and systemic circulation. Each pump has a filling chamber, the atrium, which helps to fill the ventricle, the stronger pump.

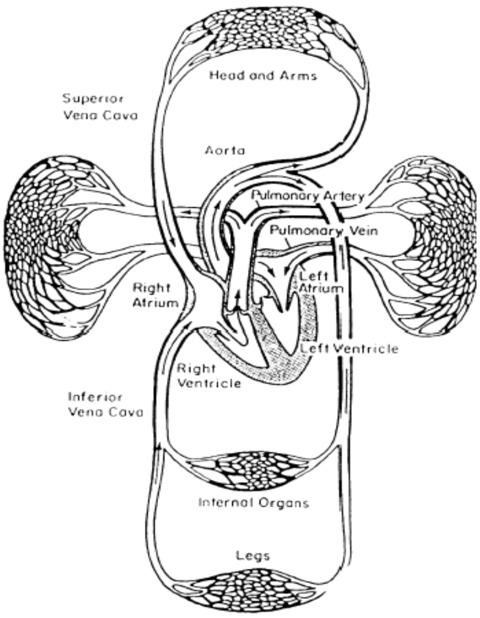
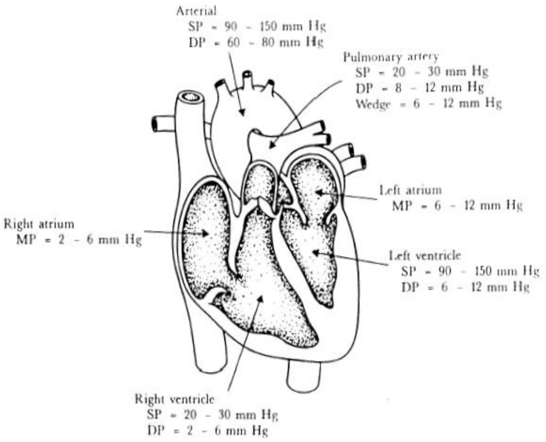


Figure 2: Schematic diagram of circulatory system

The blood flows through the various parts of heart and generates pressure. This pressure, generated by the right and left sides of the heart differs somewhat in shape and in amplitude (see Figure 3).

Cardiac contraction is caused by electric stimulation of the cardiac muscle, an electric impulse is generated by specialized cells located in the right atrium and quickly spreads over both atria. Then with specialized conduction system, conduction spreads over the interior and propagates in both ventricles. This impulse causes mechanical contraction of both ventricles and generates ventricular pressures that force blood through the pulmonary and aortic valves into pulmonary and systemic circulation, causing pressure in each. The maximum value is called Systolic pressure, the minimum value is called diastolic pressure and then there is the Mean pressure. All pressures are usually measured in millimeters of mercury (torr).



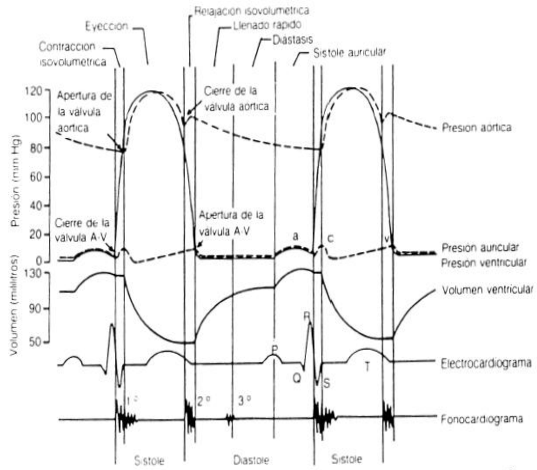


Figure 3: Typical values of circulator pressure and graph of phenomenon to monitor

How was already said, there are two types of blood pressure measurement systems, non-invasive and invasive. The most known non-invasive systems are based on the use of the sphygmomanometer. It’s used placing an inflatable cuff smoothly and snugly around the left arm, then the cuff is inflated 'till the artery is completely occluded. Listening with a stethoscope the examiner slowly releases the pressure in the cuff. As the pressure in the cuffs falls, a "whooshing" or pounding sound is heard when blood flow first starts again in the artery. The pressure at which this sound began is noted and recorded as the systolic blood pressure. The cuff pressure is further released until the sound can no longer be heard. This is recorded as the diastolic blood pressure. Automatic systems use the method by recording the oscillation superposed to the pressure signal, instead of the sounds.

Otherwise Arterial blood pressure is most accurately measured invasively through an arterial line.

Invasive arterial pressure measurement with intravascular cannulae involves direct measurement of arterial pressure by placing a cannula needle in an artery (usually radial, femoral, dorsalis pedis or brachial).

The cannula must be connected to a sterile, fluid-filled system, which is connected to an electronic pressure transducer. Historically, a number of different kinds of sensors have been used: strain gage, linear-variable differential transformer, variable inductance, variable capacitance, piezoelectric. Currently, almost all systems are based on piezoresistive sensors. The advantage of this system is that pressure is constantly monitored beat-by-beat, and a waveform (a graph of pressure against time) can be displayed [1].

The first recorded instance of the measurement of blood pressure was in 1733 by the Reverend Stephen Hales. A British veterinarian, Hales spent many years recording the blood pressures of animals. Fifteen years beforehand, he took a horse and inserted a brass pipe into an artery. This brass pipe was connected to a glass tube. Hales observed the blood in the pipe rising and concluded that this must be due to a pressure in the blood. At this time the technique was invasive and highly inappropriate for clinical use.

Figure 4: Haley measurement experiment

It was not until 1847 that human blood pressure was recorded. The method used Carl Ludwig's kymograph with catheters inserted directly into the artery. Ludwig's kymograph consisted of a U-shaped manometer tube connected to a brass pipe cannula into the artery. The manometer tube had an ivory float onto which a rod with a quill was attached.

This quill would sketch onto a rotating drum hence the name 'kymograph', 'wave writer' in Greek. However blood pressure could still only be measured by invasive means.

The lack of a non-invasive method of determining this new idea of blood pressure lead to many physicians working in this field. Once such man, Karl Vierordt, found in 1855 that with enough pressure, the arterial pulse could be obliterated. Vierordt used an inflatable cuff around the arm to constrict the artery. Etienne Jules Marey, a French physician/cinematographer, developed this idea further in 1860. His sphygmograph could accurately measure the pulse rate, but was very unreliable in determining the blood pressure. Yet this design was the first that could be used clinically was a small degree of success.

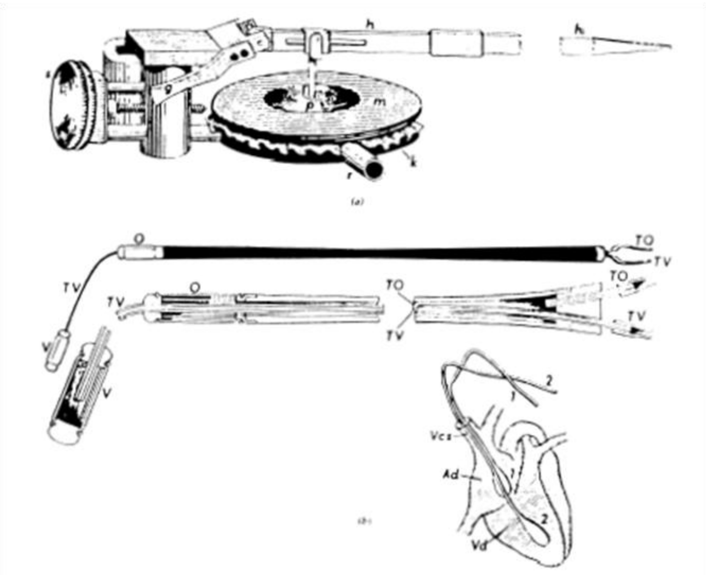


Figure 5: Marey measurement system

In 1881, Samuel Siegfried Karl Ritter von Basch invented the sphygmomanometer. His device consisted of a water-filled bag connected to a manometer. The manometer was used to determine the pressure required to obliterate the arterial pulse. Direct measurement of blood pressure by catheterisation confirmed that von Basch's design would allow a non-invasive method to measure blood pressure. Feeling for the pulse on the skin above the artery, was used to determine when the arterial pulse disappeared. However von Bacsh's design never won a keen following, many physicians of the time being sceptical of new technology, claiming that it sought to replace traditional ideas of diagnosis.

In addition, many questioned the medial usefulness of information about the blood pressure. This did not stop some from attempting to produce a more useful device. A spring-based sphygmomanometer won some support, but they were difficult to calibrate and were very unreliable when dealing with acutely ill patients. Scipione Riva-Rocci developed the mercury sphygmomanometer in 1896.

This design was the prototype of the modern mercury sphygmomanometer.

An inflatable cuff was placed over the upper arm to constrict the brachial artery. This cuff was connected to a glass manometer filled with mercury to measure the pressure exerted onto the arm. Riva-Rocci's sphygmomanometer was spotted by the American neurosurgeon Harvey Cushing while he was travelling through Italy. Seeing the potential benefit he returned to the US with the design in 1901. After the design was modified for more clinical use, the sphygmomanometer became commonplace. Cushing and George Crile were major advocates of the benefits. This sphygmomanometer could only be used to determine the systolic blood pressure. Observing the pulse disappearance via palpitation would only allow the measuring physician to observe the point when the artery was fully constricted.

Nikolai Korotkoff was the first to observe the sounds made by the constriction of the artery in 1905. Korotkoff found that there were characteristic sounds at certain points in the inflation and deflation of the cuff. These Korotkoff sounds were caused by the abnormal passage of blood through the artery, corresponding to the systolic and diastolic blood pressures. A crucial difference in Korotkoff's technique was the use of a stethoscope to listen for the sounds of blood flowing through the artery. This auscultatory method proved to be more reliable than the previous palpitation techniques and thus became the standard practice [2].

However, there are still lots of studies about invasive blood pressure measurement systems, to reach out more accuracy in pressure measurement and more safety from infections, one time caused by bad catheters, for these systems.

2.2 Invasive measurements with catheter

This measurement is called invasive because bodily entry is made. For invasive arterial blood pressure measurement an artery is cannulated. Such a system yelds blood pressures dependent upon the location of the catheter tip in the vascular system. It's particularly useful for continous determination of pressure changes at any instant in dynamic circumstances. More commonly used sites to make continous observations are the brachial and radial arteries. The femoral or other sites may be used as point of entry to sample pressures at different locations inside the arterial tree, or even the left ventricle of the heart. Entry through checks of pressure in the central veins close to heart, the right atrium, the right ventricle and the pulmonary artery. A catheter with a balloon tip carried by blood flow into smaller branches of the pulmonary artery can occlude flow in the artery from the right ventricle so that the tip of the catheter reads the pressure of the left atrium, just downstream.

These procedures are very complex and there is always concern of risk of hazard as opposed to

benefit. Invasive access to asystemic artery involves considerable handling of a patient.
The longer a catheter stays in a vessel, the more likely an associated thrombus will form for example.
Some of the recognized contraindications include poor collateral flow, occlusive arteria disease and others. In spite of well-studied potential problems, invasive blood pressure measurement is generally accepted as the gold standard of arterial pressure recording and presents the only satysfactory alternative when conventional cuff techniques are not successful.

Catheter-Tubing-Sensor System

A large variety of vascular catheter exist. Catheter materials have undergone testing to ensure the they have a minimal tendency to form blood clots on their surface. The catheter chosen maybe inserted percutaneosly over a hollow stylet into the blood vessel (Percutaneous method). Less often, entry to a vessel requires a “cutdown”, a direct exposure of the vessel after a skin incision.

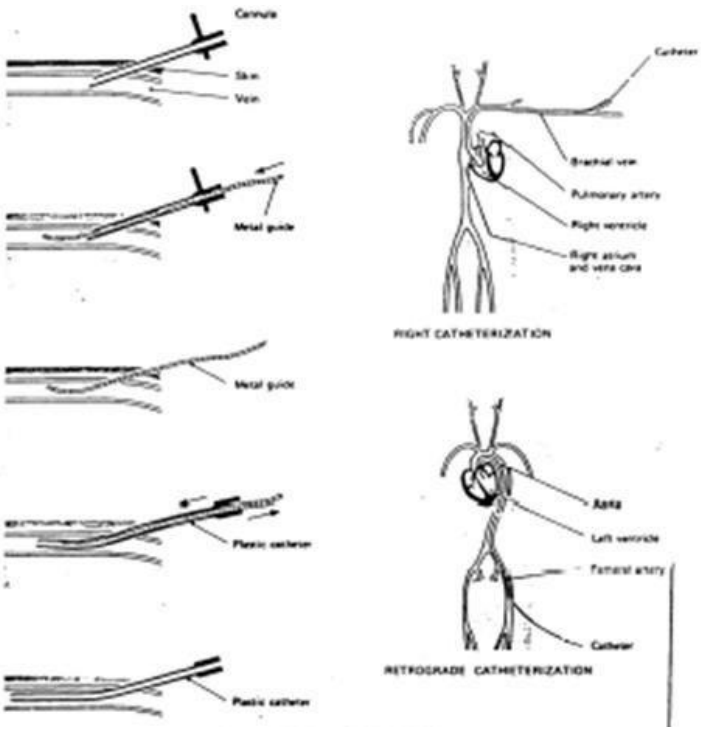


Figure 6: Example of "cutdowns"

Although pressure sensors can be located at the catheter tip, this presents problems for calibration if left in place and a clot forms near the tip of the catheter, damping the pressure signal. In addition, this catheter is very expensive.

Instead, most catheters connect to an external pressure sensor via fluid-filled low-compliance tubing. A basic system can consist of an intravascular catheter connected to a rigid fluid-filled catheter and tubing which communicates the pressure to an elastic diaphragm, the detection of which is detected electrically.

There is a direct relationship between the defection of the diaphragm and the voltage. The higher the voltage, the greater the pressure. Continuous low-rate infusion of heparinized saline is carried out to keep the catheter free from coagulations. The advent of disposable sensor kits have greatly simplified the clinical use of intravascular monitoring. This may cause more costs.

Although direct recording is considered the most accurate method, its accuracy may be limited due to the kinetic energy of the fluid in the catheter or dynamic frequency response of the measurement system. Damping and degrading the system's natural frequency, caused by trapped air bubbles, small catheters, various narrow connections, compliant and too long tubing, and too many components connected, are the two characteristic problems with a pressure sensor system [3].

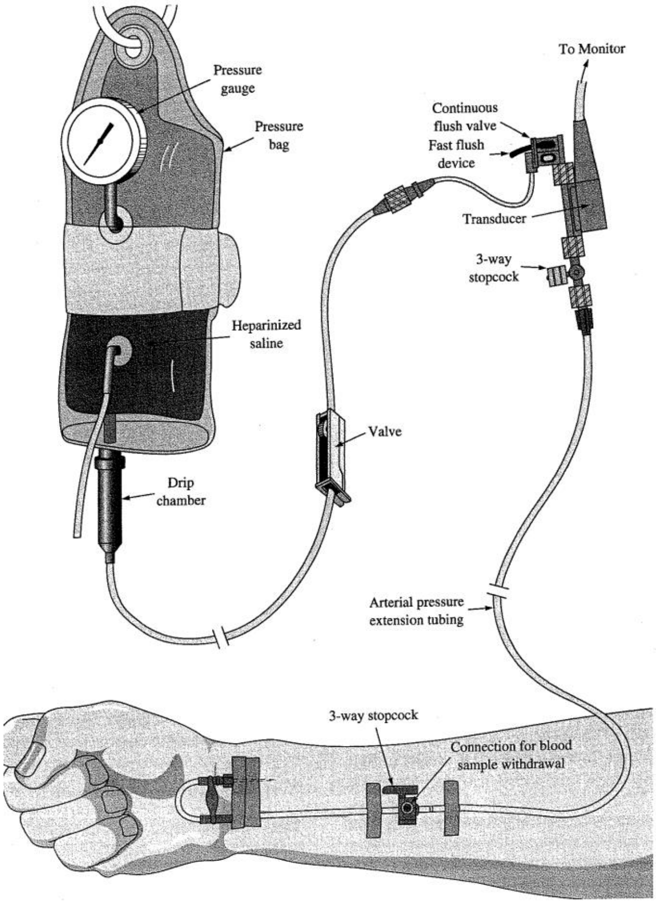


Figure 7: Schematic of an invasive measurement system[3]

Transducers

The most common type of pressure transducers consists of a diaphragm, one side of which is open to the atmosphere and the other connected to the pressure which is to be measured.
Pressure causes a proportional displacement of the diaphragm which can be measured in many ways. The most common method is to use strain gauges. A strain gauge is a device which measures deformation or strain.
A single crystal of silicon with a small amount of impurity will have an electrical resistance which changes with strain. If a silicon strain gauge is attached to the diaphragm of the pressure transducer then its resistance will change with the pressure applied to the diaphragm.
Unfortunately the resistance of silicon will also change the temperature, but the effect of this change may be reduced by attaching four strain gauges to the diaphragm, like a resistive bridge.

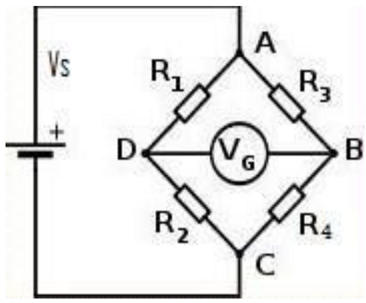


Figure 8: Diagram of the resistive bridge

These strain gauges then form the four arms of a resistance bridge. By placing two of the strain gauges tangentially and close to the centre of the diaphragm, where the strain is positive and hence their resistance will increase with applied pressure, and the other two radially and close to the periphery, where the strain is negative and hence their resistance will decrease with applied pressure, the resistance bridge will be unbalanced by a change in pressure. However, if the temperature changes, all four resistances will change by the same percentage and this will not change the output from the resistance bridge.
If $R_1 = R_4 = R_0(1+x)$ and $R_2 = R_3 = R_0(1-x)$ where R_0 is the resistance when the strain is zero, we can find the output of the system like:
$V_g = V_s * x$ being x the relative change in the resistance, due to the strain.
The complete transducer may have a dome, which can be unscrewed for cleaning and sterilization but it more likely that the transducer will now be a disposable type.

The next figure shows how the transducer can be connected for an arterial pressure measurement:

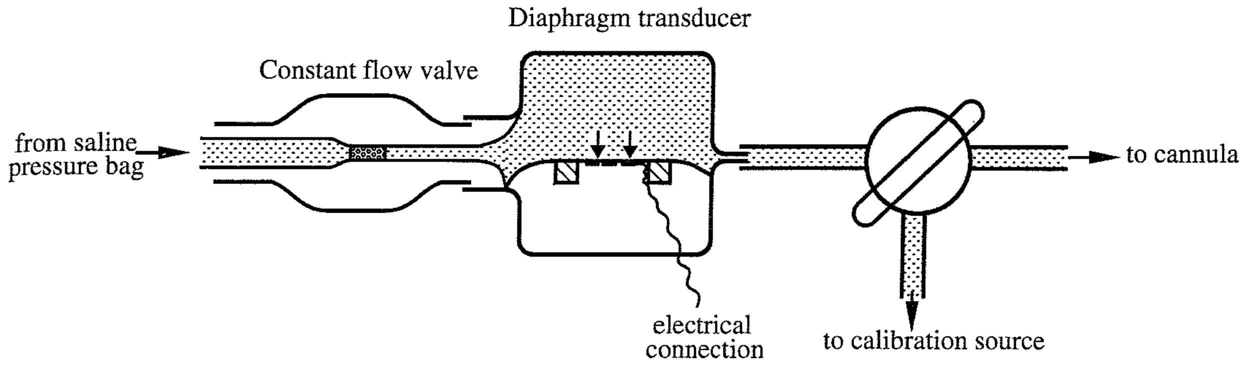


Figure 9: A disposable pressure transducer connected to an arterial catheter/cannula[4]

It's very important to remove all bubbles of air, if air bubbles remain then false pressure readings will be obtained and the frequency response of the transducer will be reduced.

2.3 Dynamic characteristic of catheter-sensor system

The addition of a flexible fluid-filled catheter to the relatively rigid transducer introduces severe problems associated with the dynamic performance of the system. We'll use a simple lumped parameter model, analogous to an R-L-C circuit, to explore the dynamic performance problems.
Next figure shows the mechanical and the electrical models of transducer and catheter:

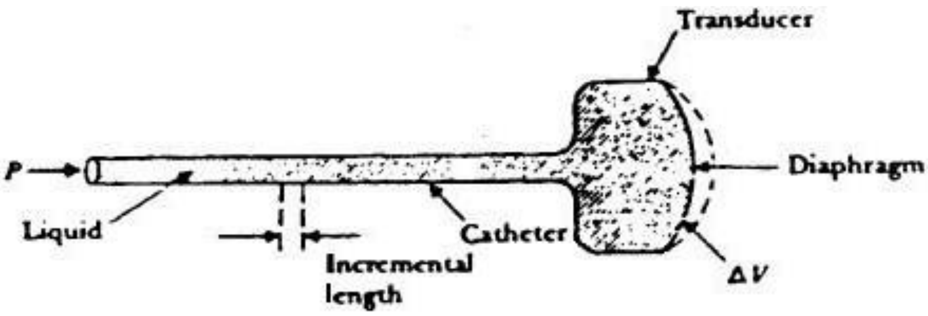


Figure 10: Mechanical model of transducer and catheter

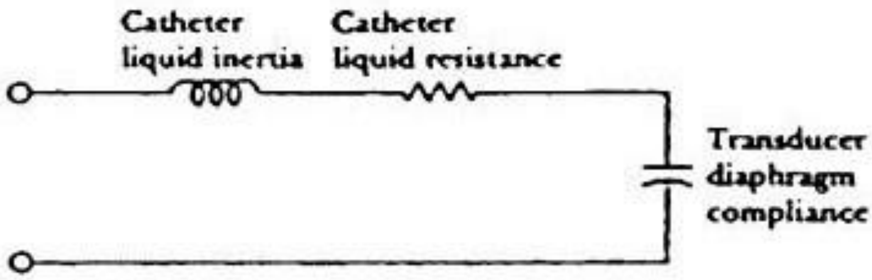


Figure 11: Electrical model of transducer and catheter

A pressure change at the end of the catheter will cause fluid to move through the catheter and displace the diaphragm of the transducer. The hydraulic properties that are important are inertance, resistance, and compliance, the electrical analogues of which are inductance, resistance and capacitance. These properties are due to inertia, friction and elasticity within the system. In theory we have to divide in inertance, resistance and compliance for the catheter, the transducer and diaphragm (that is mainly a capacitance), the dominant elements are: inertance and resistance for the column of the liquid column in the catheter and compliance in the diaphragm.

The resistance Rc is due to shear forces within the liquid flow through the catheter, and is given by

$R_c = \frac{\Delta P}{F}$ $[Pa\,s\,m^{-3}]$ where ΔP is the pressure difference and F is the flow rate.

Poiseuille's law relates the pressure differential and flow rate to the viscosity and size of the tube:

$$\frac{F}{\Delta P} = \frac{\pi r^4}{8\eta L} \Rightarrow R_c = \frac{8\eta L}{\pi r^4}$$

The inertance L_c is the ratio of the pressure differential to rate of change of flow:

$L_c = \frac{\Delta P}{(\frac{dF}{dT})}$ where a is the acceleration and A is the cross-sectional area of the catheter.

The compliance C_d of the transducer diaphragm is defined as

$C_d = \frac{\Delta V}{\Delta P} = \frac{1}{E_d}$ where E_d is the volume modulus of elasticity of the transducer diaphragm.

If we now apply a sinusoidal input voltage v_i to the model (analogous to a varying pressure), we can calculate the output voltage across C_d (analogous to the measured pressure).

We do this by summing the voltages around the circuit to give

$$\frac{d^2 v_0}{dt^2} + \frac{L_c}{R_c}\frac{dv_0}{dt} + \frac{1}{L_c C_d} v_0 = \frac{1}{L_c C_d} v_i$$

and compare this to the standard form

$$\ddot{x} + 2\omega_0\zeta\dot{x} + \omega_0^2 x = y$$

where the damping factor $\zeta = \frac{R_c}{2}(C_d/L_c)^{1/2}$ and the natural frequency $\omega_0 = (L_c C_d)^{1/2}$

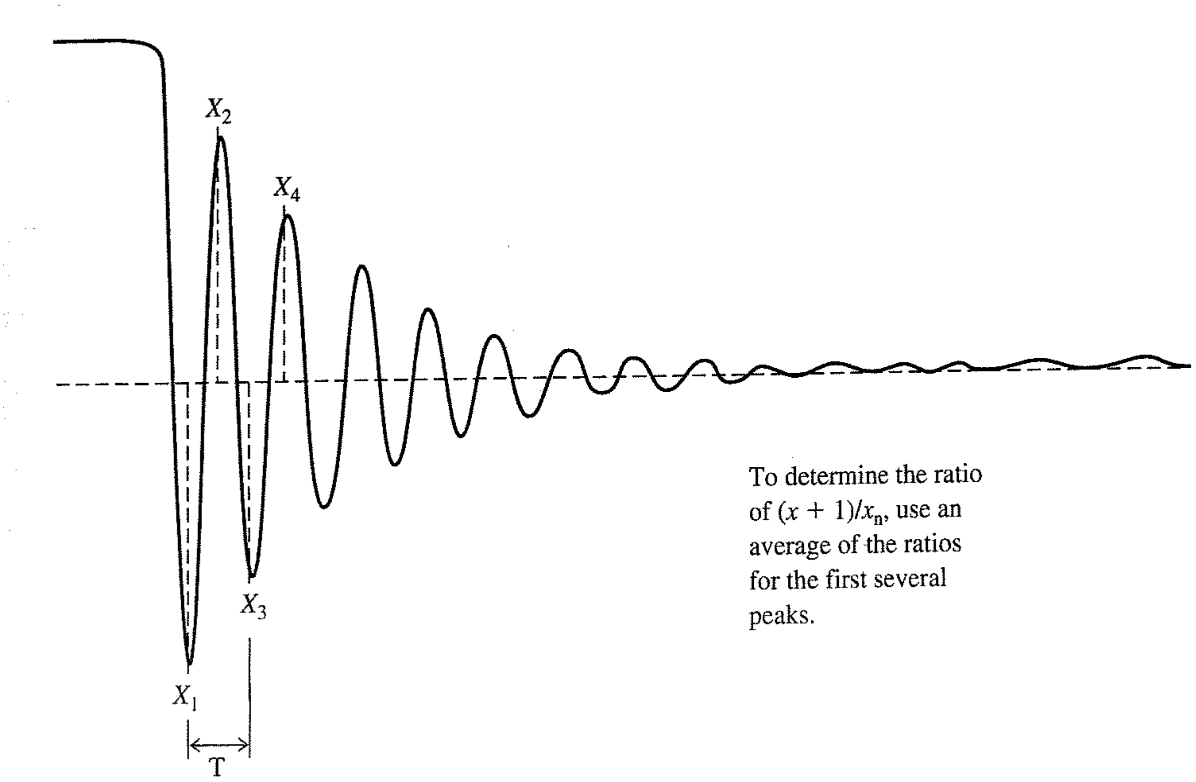


Figure 12: Step response of the catheter sensor system for a sinusoidal voltage stimulus[4]

Although exists another method to calculate damping factor and the natural frequency, it's possible to look at the peaks of the transitory and using this formulas arrive at the final result:

$$\zeta = \sqrt{\frac{ln^2(\frac{X_{n+1}}{X_n})}{\pi^2 + ln^2(\frac{X_{n+1}}{X_n})}} \quad \text{and} \quad \omega_0 = \frac{1}{T\sqrt{(1-\zeta^2)}}$$

with this parameters it will possible to calculate the frequency response.

It is usually desirable for the damping factor to be on the order of 0,7 for a frequency response of 20 Hz. This number provides critical damping. If the number is very much less, then the system is subcritically damped and may ring (providing unnaturally high systolic reading resulting from a sharpering of the waveform caused by excess high-frequency components).
Similarly, an undercritically damped system, results in attenuation of high- frequency components, causing an unnatural low systolic reading.

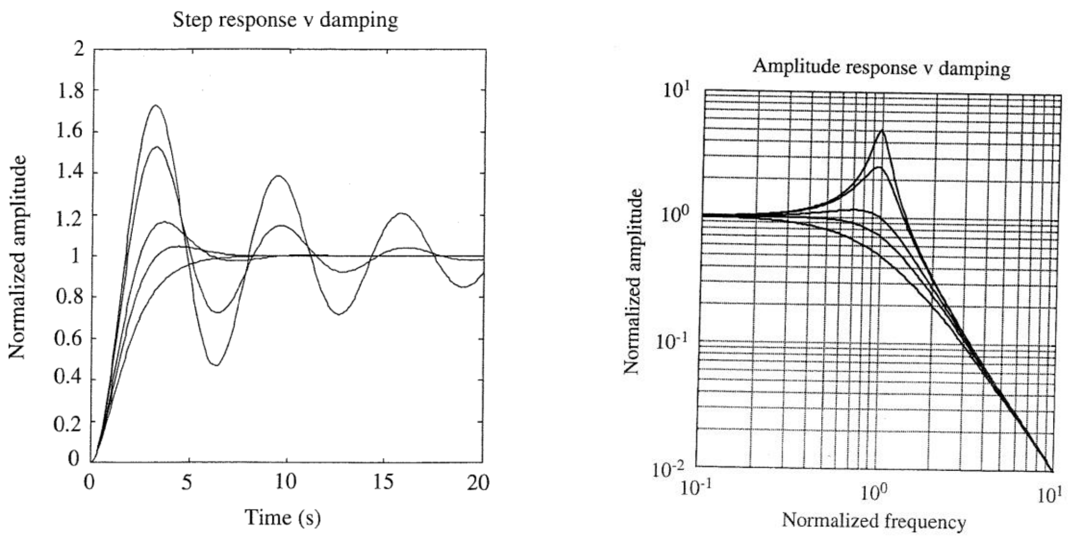


Figure 13: a) Step response with different damping-lower damping gives more over shoot
b) Frequency response with different damping-lower damping gives a high resonant peak[4]

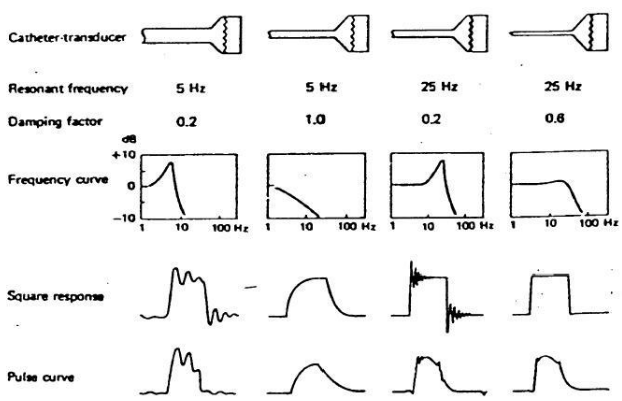


Figure14: Effect of the system response in the measurement of a square pressure pulse and of a typical arterial blood pressure pulse

Moreover, if we introduce a bubble of air (with high compliance) into the catheter, we can model it by adding an additional capacitance C_b in parallel with C_d .

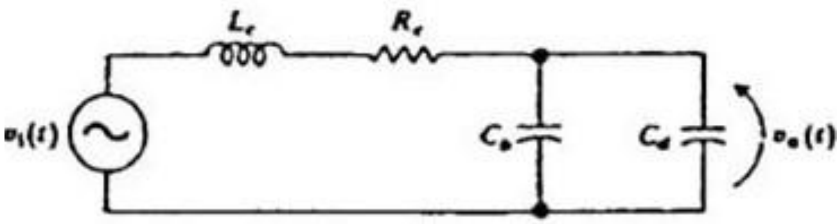


Figure 15: Electrical model introducing bubbles

The compliance C_b of a bubble is given by

$$C_b = \frac{\Delta V}{\Delta P} * (volume\ of\ air\ bubble)$$

The introduction of bubbles of air will have a damping effect reducing the high-frequency components and given a lower frequency response [4].

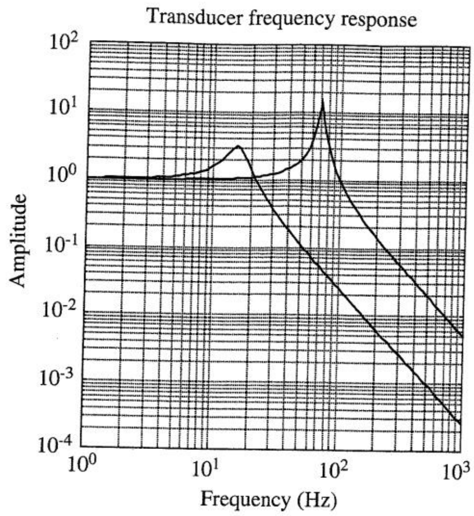


Figure 16: Frequency response for transducer and catheter. The lower frequency response is with a bubble in the catheter[4]

3 Analysis of the problem

3.1 Specifications

How already said in the introduction, the objective of this work is not to produce a real invasive pressure measurement system, but to realize an experimental board to reproduce the dynamic behavior of the arterial pressure, to give a didactical tool to be used in the remote laboratory of the UPC, with connection with the remote board, that is used in the laboratory of Instrumentation and Bioengineering. This was a little advantage, because permitted to adapt the system to the components, most of those were already in the laboratory, only buying some electronics components, like the sensors, the amplifiers, etc. To make this parallel board we have exploited of some little changes to reproduce the single components of the system:

- First, the substitution of the real commercial catheter for invasive measurements, with simple plastic tubes, to make the measures. One is made of a strong plastic, but at the same time is flexible, the other is made of a more soft plastic, with a larger diameter and, moreover, longer. The idea is to generate different measurement results and also compare the differences between the tubes, making measures with air bubble or no.

- The second change was about the substitution of the blood with other fluids. Initially was thought to change blood with water, the more achievable fluid in the world. All measures were made with the water and have reproduced satisfactorily the dynamic characteristic of the invasive system. But, there are some problems with water, because water can leave residues of limestone, for example, in the tube and then ruine it with the help of the time. Then, we need a liquid which does not degenerate with time nor contaminate. A good candidate would be the “Oraldine” for washing teeth.

- For the sensor conditioning we have used instrumentation amplifiers, to condition the differential voltage signal in output from the sensors and to control the gain with a simple resistance. A constant power supply to the sensors was guaranteed by a voltage regulator, connected with a trimmer and a polarized capacitor.

Static characterization

To reproduce the static behavior of the system and verify the linearity of this one, we will use a linear stepper motor, connected to a piston. Both the motor and the piston have an important feature; they can move in all the two directions. Making these it will be possible to reproduce the positive and negative pressure and study the behavior with the two different stimulus. Then the idea will be to connect the piston to a net of catheters connected to the sensor, using a little plastic tube, also to put water in the piston, and reproduce a good background of work. This, maybe, will be very useful to verify the correct working of the sensors, viewing, making like a ratio of two sensor's responses and proving that it will be a straight line.

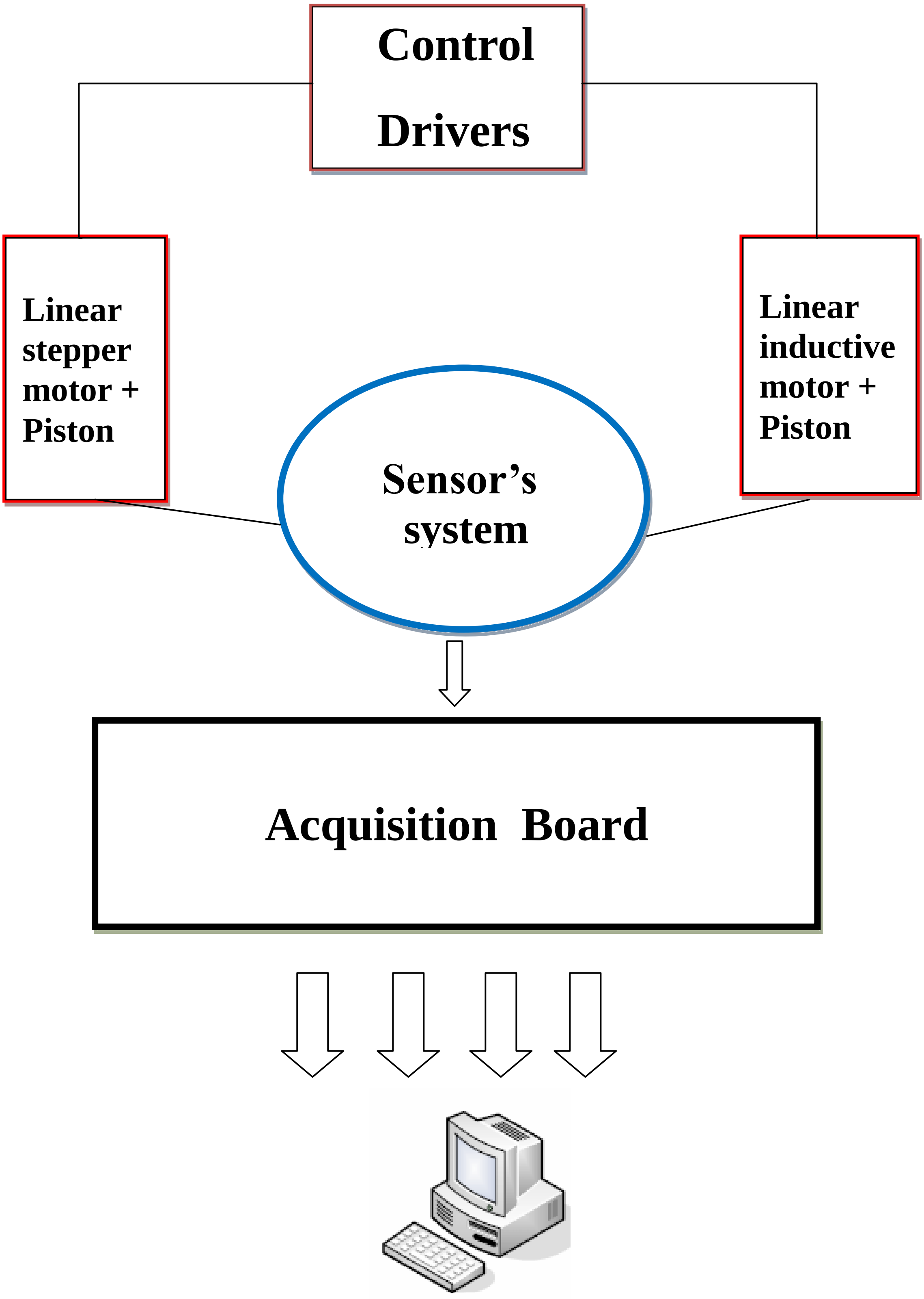
Dynamic characterization

The first step to make the dynamic characterization of the system will be to put water in a tube circuit connected to the sensors, and at the other end of the circuit we'll put a syringe with water too. This well reproduced the dynamic behavior of the heartbeat, and we will be able to see, using an oscilloscope, for the first time the damping signal, produced hitting softly the syringe.

The second step to reproduce the dynamic comportment of the system will be to put a linear inductive motor in place of the syringe. This motor has the characteristic of producing an impulsive signal of almost 100 ms. This was very important, because, using this load, we could reproduce the dynamic response of the heartbeat.

Like in the static characterization this motor will be connected to a piston, with the same features of the previous one. In fact, also here, both the motor and the piston have the same important characteristic, the possibility of moving in the two directions, forward and backwards. This will permit to take the best direction of measurement and if the pressure was negative, it will be possible to modify this, introducing a simple multiplying factor "-1" in the Labview software.

3.2 Block Diagram



4 Design of the solutions

4.1 Control and acquisition system

To be able of making the measures, remotely connected with the PC, we have used an experimental mixed-mode board, already present in the laboratory, that permitted to record the static and dynamic responses only putting our board in its connectors. This board was thought in the laboratory of Instrumentation and Bioengineering to guarantee a remote tool for didactical purposes.

The main field of applications of this board is the Sensor and Signal Conditioning subject, but is also used in Electronic instrumentation subject. The topics to cover will be thus related to understanding different measurement techniques, signal conditioning and acquisition, error compensation, systems calibration and data processing.

Hardware structure

The modular laboratory platform is based on a set of printed circuit boards. Each laboratory activity is implemented on a separate application circuit board (10cm x10 cm) which is connected to a

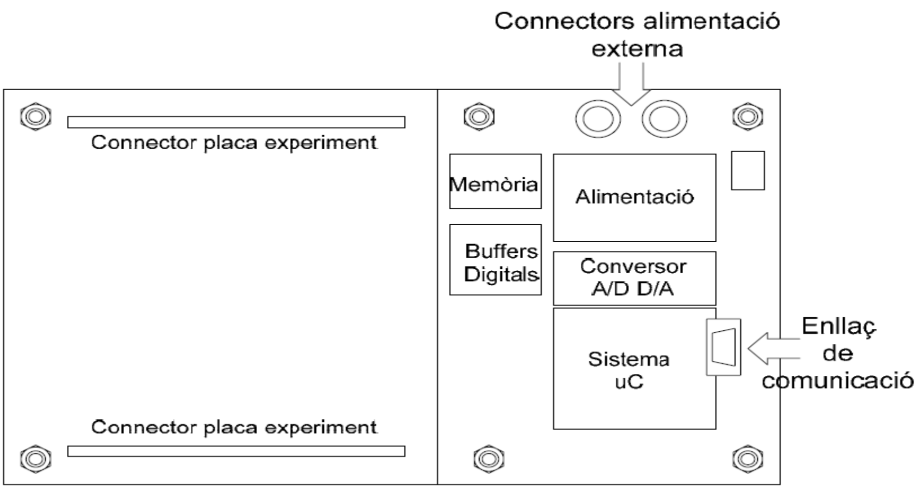


Figure 17: Block diagram of the mixed-mode board[6]

motherboard from which power supply, analog and digital inputs and some of the adjustments are provided. The motherboards are implemented on Europa format (16 cm x 10 cm) cards. This board includes A/D and D/A converters and an embedded controller which provides TCP-IP access to the acquired signals.

An embedded microprocessor, based in Rabbit RCM2200 controls the device's board via SPI serial communication and allows to connect the board via ethernet using a RJ-45 port to the laboratory server, who runs the Labview applications and displays the interested results in-situ or remotely.

The motherboard PCB includes the following devices:

- Rabbit Semiconductor RCM2200 microprocessor module.

- AD7734 four channels for A/D converter and AD5327 four channels for D/A converter.

- ST Microelectronics M95256 expansion memory (256 kb) with pin-out compatible with larger memories.

- AD780 voltage reference (2.5 V) for A/D and D/A.

- LT1964 dc to dc converter, LT1521 positive linear regulator and LT1964 negative linear regulator.

From a single 5 VDC power input, $\pm 5\,V$ are provided to the experiment board and 5 V to the auxiliary digital circuits, being this power supplies enabled by the microcontroller.

Moreover the board can provide 8 control bits.[5]

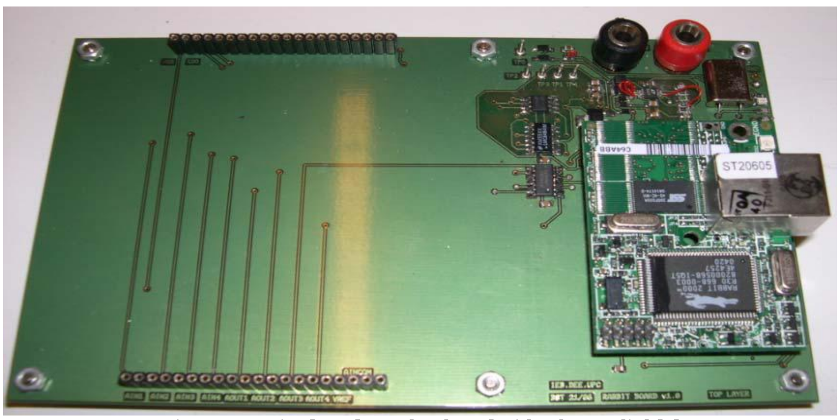


Figure 18: Mixed-mode mother board with ethernet link[6]

Software structure

The microcontroller firmware implements a command parser the accepts SCPI command via ethernet link which configure the acquisition channels and output signals, acquires output signals in several modes and sends the desired information to the TCP/IP client. At the same time, the personal computer hosts the Labview server, running specific software for each experiment processes the data transferred and control the communication. This firmware was developed with Dynamic C 9.10.

To control the Rabbit was developed a firmware that could support all the instructions to control all the experiments. Data format, sent between µC and PC, will be strings of characters. So, the firmware will have to detect what is wanted to be done for each case through Ethernet connection and to return the value, if necessary.

These are the basic functions used to control the µC:

*IDN?: system identification. Show into the monitor the system configuration.

*RST: make the reset of the A/D and D/A converters and recover the default settings.

MEAS?: this indicates the operation of measurement. This instruction has three variants:

SX: S is for *single,* X is the acquisition channel;
CX: C for continuous.

MOVEDSXXX: generate the control signal for a stepper motor driver, for the applications where is necessary.

D: direction of rotation: forward or backward;
S: specify if motor steps are complete or no;
XXX: motor step’s number.

FAST: string to enable fast measures, needed for taking the dynamic characteristic. [6]

Labview Driver

Labview drivers are specific programs to send instructions correctly, depending of what's requested. To establish the connection with the Rabbit through TCP/IP were fixed two variables for IP address and the used Port. This permits to connect the driver Labview to the measurement board.

Data	Description
Direcció IP abc	Programmed IP address for the Rabbit.
Port Remot U16	Port to enter in the Rabbit with TCP/IP communication.

Figure 19: Shared variables for the drivers

Control drivers used in Labview are:

Simple measure

Using the variable "Canal Mesura" it's possible to fix the channel for the measurement

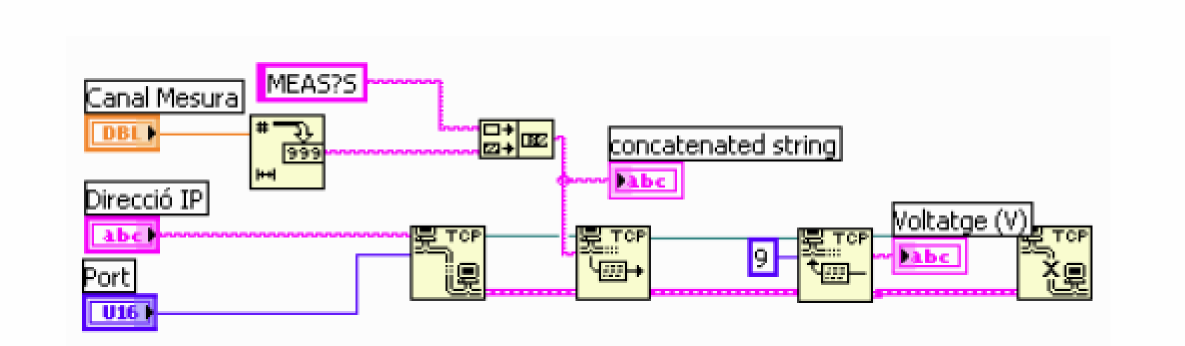


Figure 20: Diver for simple measure

Into the software there are basic blocks for the TCP/IP communication with Labview. Blocks are:

- TCP Open Connection: to open a TCP/IP connection;
- TCP Write: to write the string that identifies the instruction for the Rabbit;
- TCP Read: to read the information, defining reading bytes number;
- TCP Close connection: to close the TCP/IP connection.

Motor controller

For this driver are used the TCP blocks to control the connection. The other variables represent the direction ("Turn (0 o 1)"), the complete step ("Step (0 o 1)") and the number of steps ("Clock Number"). Finally the string "MOVE" to control the Rabbit. [6]

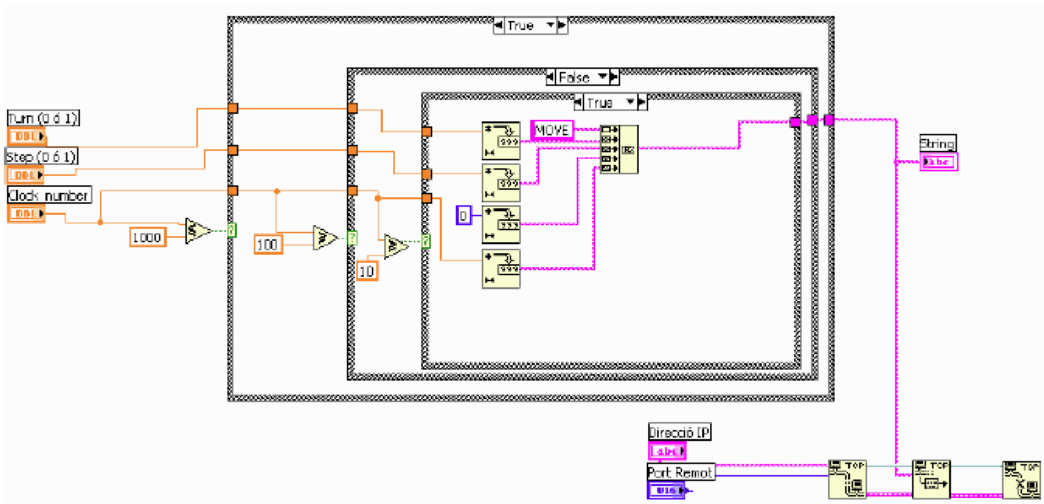


Figure 21: Driver for controlling motor

Fast measures' driver

A problem with the simple measurement driver is the fact that it can take only a few measures per second. To acquire the catheter transitory response, we need to acquire signals in the range of milliseconds.

This driver utilizes the string FAST to control the Rabbit. It's based on a for cycle that repeated 1600 times the measure to realize, through a feedback node , an n-dim array that is after divided in four 1-dim arrays that represents the four measurement channels. Then converts the strings in Output of the four arrays in number, and then converts the number in a 16-bit integer, that is the measure result.

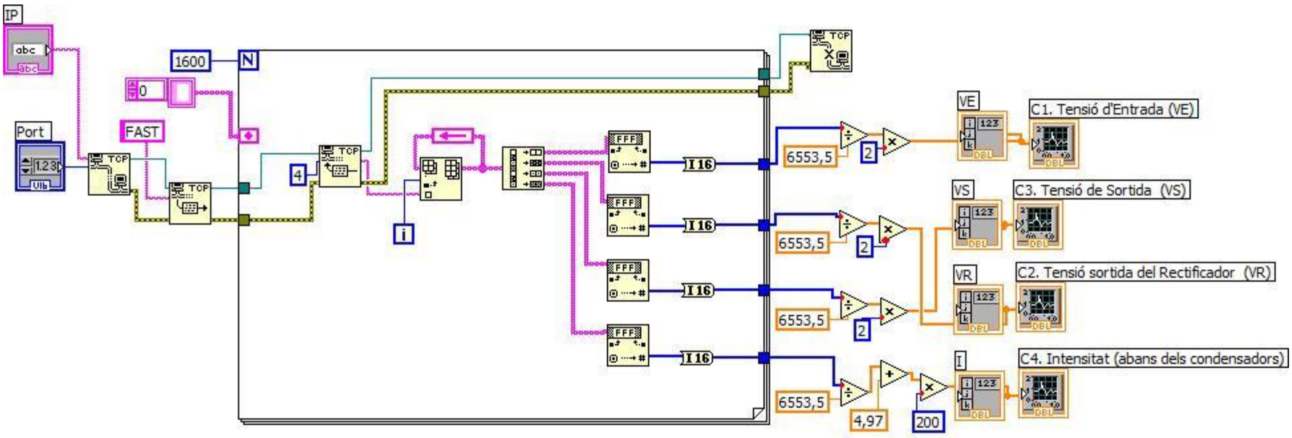


Figure 22: Driver MeasureFastLV71

4.2 System's components

There are three main parts of this system, the first are sensors and their conditioning circuits, the second are the actuators for the static characterization with stepper motor, piston and driver. Finally the third is for the dynamic characterization with an inductive linear motor, the same piston and the driver.

Sensors + Amplifiers

To take the pressure measures we have chosen the 26PCCFA6D sensors from Honeywell. This is a piezoresistive sensors that can measure pressure in a range of 0-15 psi (0-775,9 mmHg) with a power supply of 0-16 Vdc (nominal 10 Vdc).
This sensor also presents the possibility of a good temperature compensation and calibration, against a low price. Moreover, this sensor was chosen for its good behavior on pressure measurements with liquids.

Figure 23: Sensor 26PCCFA6D

To guarantee an independent and stabilized power supply of 4 Vdc to the sensor, integrated regulator TPS7101Q was used.

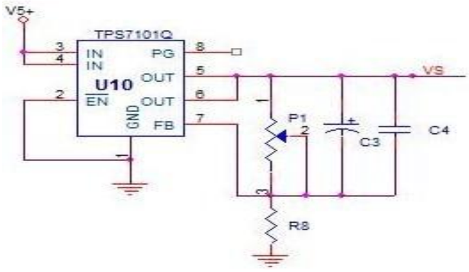


Figure 24: Voltage Regulator

Because of the Weathstone Bridge configuration of the sensor, and then the differential output, for the conditioning circuit we've used integrated instrumentation amplifiers AD620, from Analog Devices.

Those amplifiers permit to regulate the gain, only changing a single resistance, called Rg. Because of the 6,67 $\frac{mV}{psi}$ output of the sensor with a power supply of 10 Vdc and considering the fact that we feed this sensor with a voltage of 4 Vdc, it was considered an output of about 3,35 $\frac{mV}{psi}$, hence 0,065 $\frac{mV}{mmHg}$ (considering 1 psi = 51,7 mmHg). Because our full-scale pressure was of 200 mmHg, were accounted a full-scale of about 13 mV for our measure.

Finally, we wanted to reach out at the output of the measure board a voltage of about 2 V, then, in according with the component's availability of the laboratory, was established a gain of about 150, using a resistance of 332 Ω

The next figure shows the schematic of connections between sensors and amplifiers. The Vout pins are the differential output from the sensors:

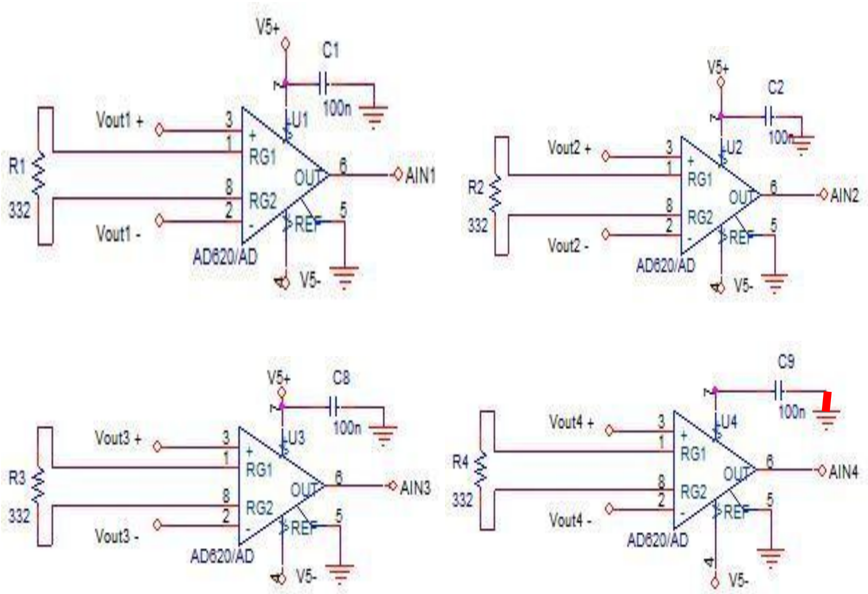


Figure 25: Schematic of conditioning circuit

Stepper motor + driver + piston

1. Motor driver

Furnished by RS, the necessary inputs are *h/f* to control if steps are full or half, *dir* to know if we go forward or backward and *cki* to count the number of steps.

This driver builds phases for controlling the engine through the outputs PHA, PHB, PHC and PHD, which are directly connected to the motor. The voltage of the driver is 12V, and it’s the same for the motor, which is supplied by the output driver VM. This because the motherboard power supply is not enough for the power needs of the motor. Then we should provide extra power supply to the experiment board. [6]

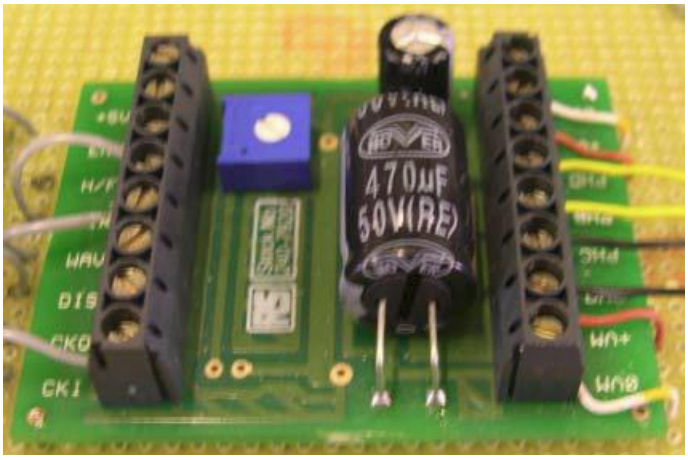


Figure 26: Step by step motor driver

2. Step by step motor

Step by step motor is a linear actuator from RS (model L92121-P2), of 7,3 N of maximum strength. Power supply is of 12 Vdc and is composed of 4 inductors.

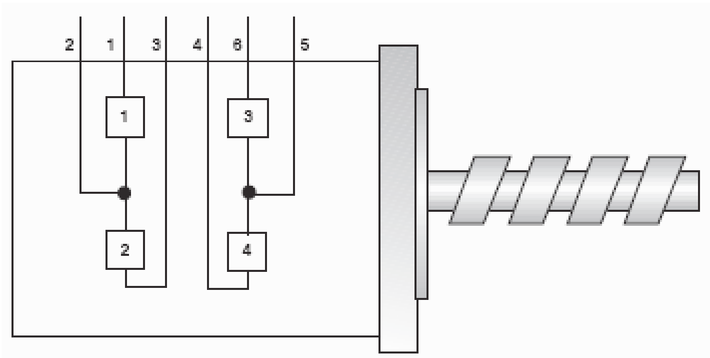


Figure 27: Internal composition of the motor

The relationship between colors and the numbers that appears in the next figure are:

Number	Colour	Connection
1	Yellow	PHA
2	Red	VM
3	Orange	PHB
4	Black	PHC
5	Green	VM
6	Brown	PHD

Figure 28: Connections of the motor with the driver

The displacement of the motor shaft for each step is of 0,0254 mm.

3. Piston

The piston used is of the manufacturer SMC, CD85 series, and is dual purpose, allowing the displacement of the piston in both directions. The main features are:

- Maximum pressure: 10 bar.
- Displacement range: 50 mm.

Figure 29: View of pneumatic piston

Although the piston will not move for the effect of the pressure, but because the piston is in agreement with the axis of the linear stepper motor.

In this way we move the piston back and forth through advancing the motor, and therefore produce changes in pressure of the piston cavity. The pressure sensors will measure the pressure on the cavity. [6]

Solenoid + driver + piston

1. Driver

The driver used to control the solenoid is the DRV102 of Texas Instruments. The next figure shows the schematic of connections of this driver. For the power supply we've used the same VM used for the linear step by step motor (12 V), furnished at pin 5.

For the Input (pin 1) was used the PE5 bit, available with the remote board. At pin 2 and 3 are connected one capacitor of 100 nF and a resistor of 91 kΩ, the first for the delay adjust and the second for the adjustment of the duty-cycle.

Pin 7 is available for connecting a LED with a resistor of 5 kΩ, to provide an advisor of over and under-current status. Pin 4 is the ground and finally pin 6 is the output of the driver, where is connected the load. We have also put two diodes, the first to maintain the hold force during PWM (pulse-width modulation) operation and the second because is necessary when is driving an inductive load.

The driver provides a sudden pulse when the control bit is set on. After a programmable period, it switches to a PWM mode to reduce the power consumption.

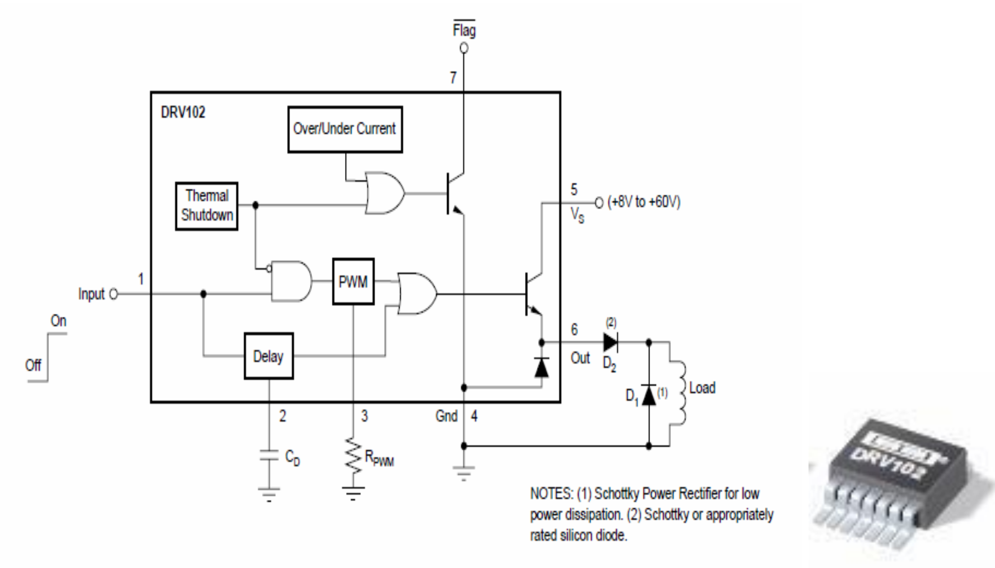


Figure 30: DRV102 and connection diagram

2. Solenoid

To reproduce the heartbeat, hence an impulse, we have used a GHUZ04M20D01 series solenoid, produced by Emessem Solenoid Company. With the power supply stored by the driver, this solenoid provides a stroke of 10 mm, that produces, commanded by the DRV102, an impulse of about 100 ms.

Moreover, this solenoid is very versatile for various operations, and is suitable for mounting and operation in any attitude.

The next figure shows an image of the used solenoid:

Figure 31: The inductive load

3. Piston

The piston used for this system is the same, already described, used for the static characterization.

4.3 The testing board

The first measurements to test the system were done with a testing board, realized in a prototype board, with components already available in the laboratory. The board presents a little system formed by the connections for the drivers and sensors, two instrumentation amplifiers AD620 and the integrated voltage regulator TPS7101Q with his conditioning circuit.

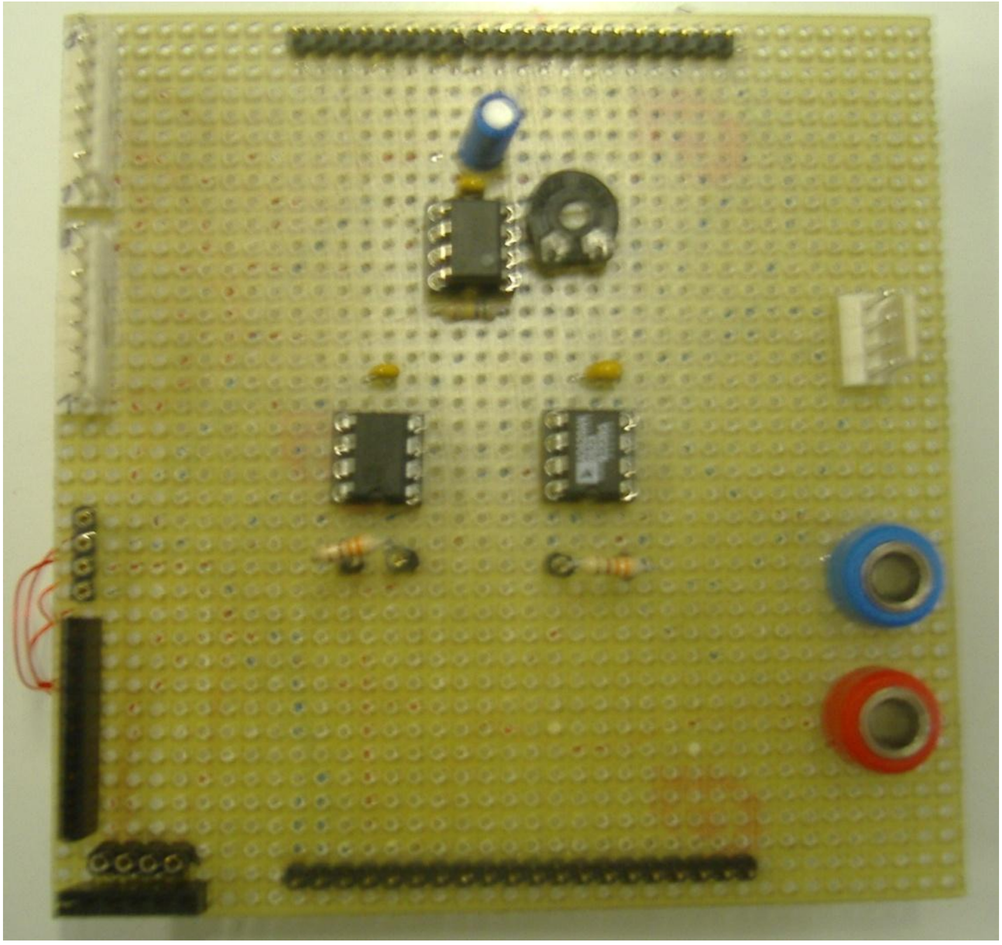


Figure 32: The testing board

This board permitted to start the study of the final board and to begin the characterization of the catheter system, using the inductive motor to begin to verify the first results.

To test the chosen inductive motor and to verify the good working of the driver DRV102, that had to produce an impulse of about 100 ms. The next picture shows the conditioning circuit for the driver implemented in a prototype board.

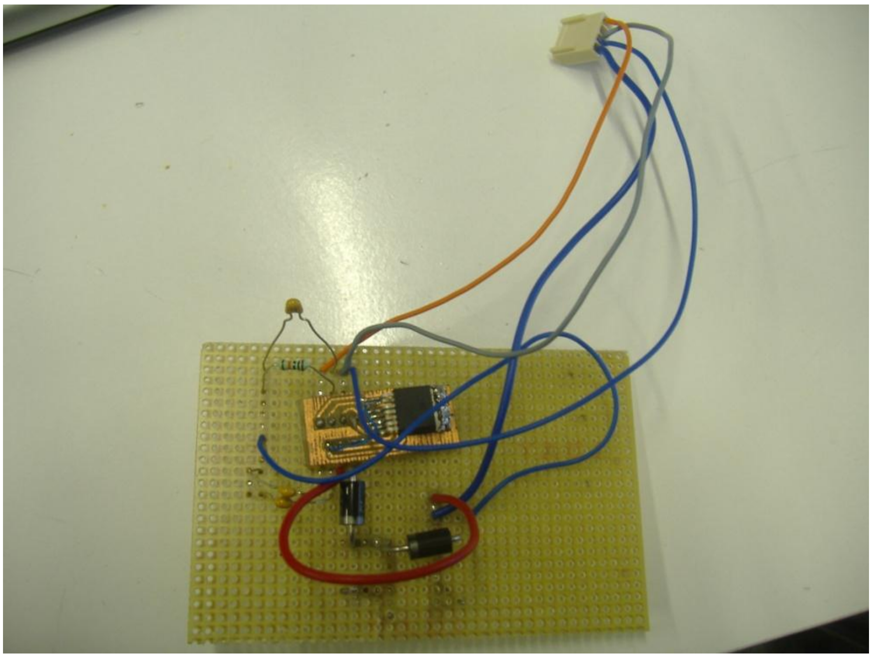


Figure 33: Conditioning circuit for DRV102

The first step was to use the system only with air; the expected results were to find a very low dynamic response, due to the strong absorbing capacity of the air. The rise time is about 30 ms.

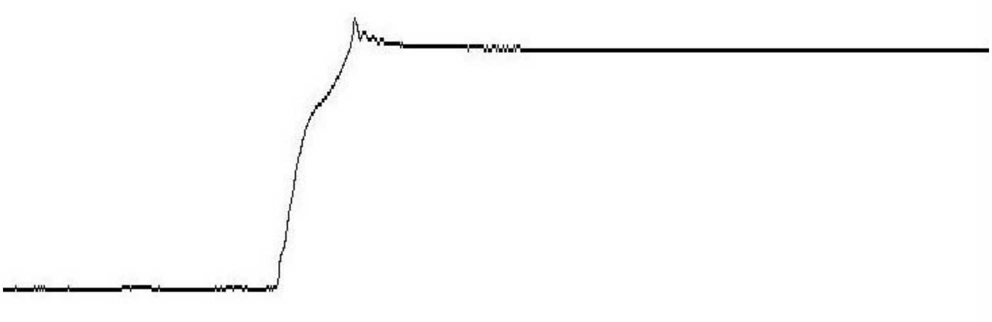


Figure 34: Measurement with air

The second step was to introduce water in the catheter system. First was used a syringe at the end of the to give the first manually-made impulse, hitting the head of the syringe with a dry stroke. Making this was noticed the first good second order system response with a transitory of about 200 ms.

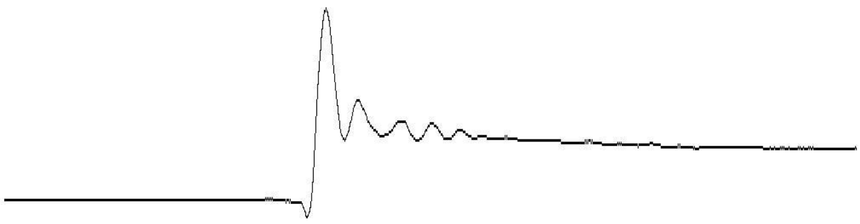


Figure 35: Measurement with water and syringe

Also with the inductive load was registered a good second order system response of about 200 ms of transitory.

This permitted to us to start studying the final circuit board with sure results and with the initial conviction of the probably good behavior of the system.

All the measures are made with an oscilloscope with an USB output, that permitted to us to save our results.

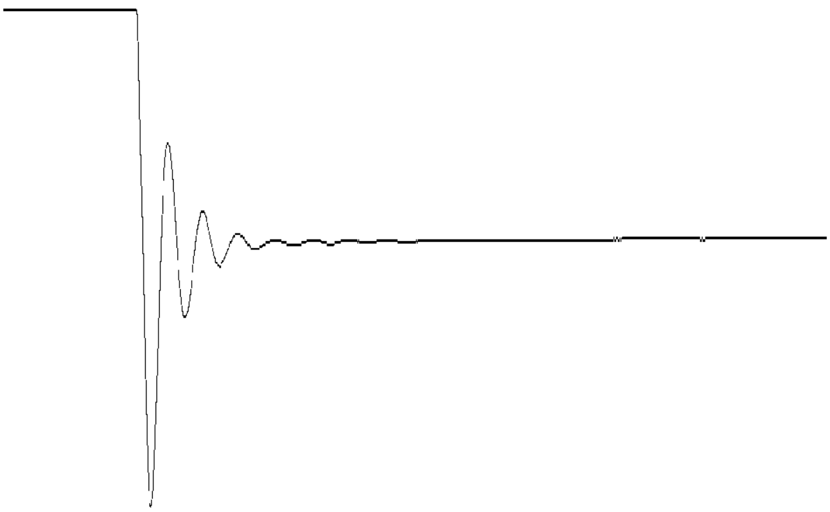


Figure 36: Measurement with water and inductive motor

5. Realization of the system

5.1 Board development

For the physical realization of the board, the work was divided in two parts, first designing the schematic, then designing the layout of the circuit board.

For the design of the schematic, implemented with Capture Cis from Orcad family, were developed all the parts divided, giving the voltage reference for each output and input.

To connect the board to the remote measurement system we have used two connectors of 20 pins. One to connect the part of positive and negative power supply and of the 8 bits with the circuit (J1), the other to connect the output of the amplifiers to the reading circuit of the measurement board (J2). Other connectors were used to connect the sensor to the circuit and to connect the motors to the power supply VM and to drivers.

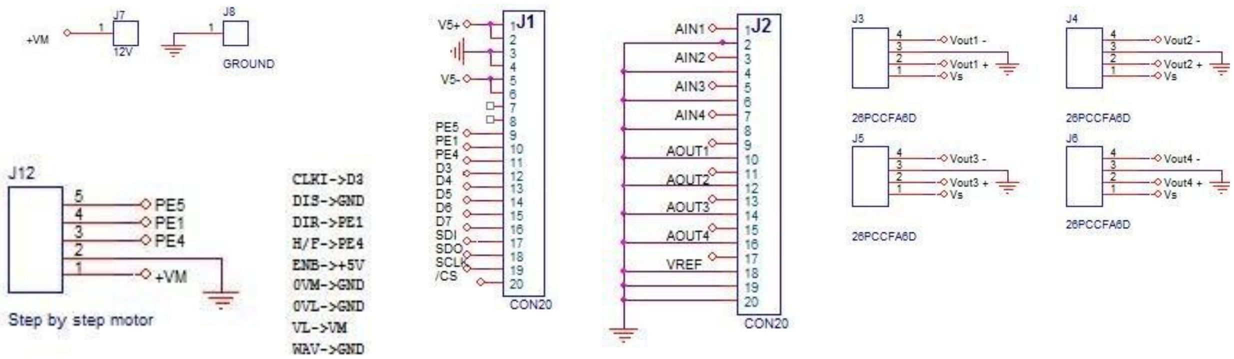


Figure 37: Schematic of the connectors

To stabilize the power supply given to the sensors, was used the circuit with TPS7101Q already described in the previous chapter, with the insertion of a trimmer (P1) of about 12,5 kΩ, a polarized capacitor of 22 µF, a capacitance of 1microF and a resistor of 560 kΩ. The output Vs of the circuit goes directly to supply the sensors.

For the conditioning of the signal in output from the sensors, like already said, was used for each one, an instrumentation amplifier AD620 produced by Analog Devices, ensuring, with a resistor of 332 Ω, a gain of 150 to read in output a full-scale voltage of about 2 V.

The power supply of the instrumentation amplifiers is ± 5 V, furnished by the measurement board.

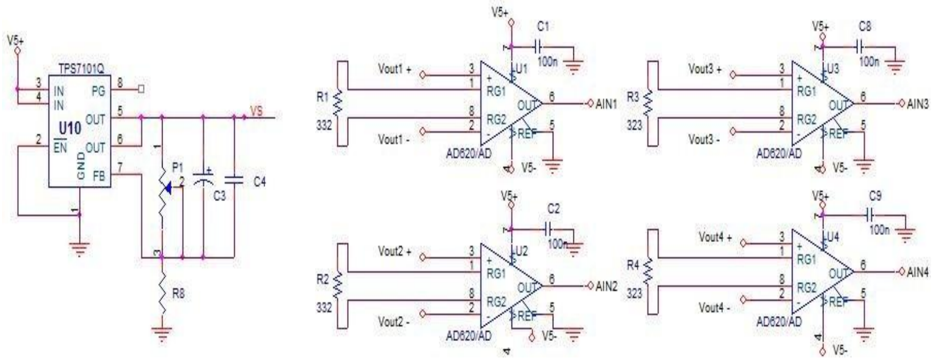


Figure 38: Schematic of the circuit for TPS7101Q and AD620

The driver for linear motor used for the static characterization is the component distributed by RS, described in the previous chapter. In this board it was necessary to connect this driver to the bits to drive the motor, these bits are PE5 for the clock, PE1 for the direction and PE4 to say if it's needed an half or full step.

To drive the inductive motor was used the driver DRV102 produced by Texas Instruments. A part in Capture Cis was adapted to reproduce the original pin setting in the schematic. For the PWM output was used a resistor of 91 kΩ, for the Delay Adjust was used a capacitance of 100 nF, which gives an pulse width of 91 ms.

To give the clock in the input pin of the driver was used the bit D3, given by the measurement board.

The LED, in output from the Flag pin, is connected to the power supply of 5V by a resistor of 5 kΩ.

Moreover, to guarantee the power supply VM to the drivers two Banana connectors were used, to add an independent power supply that isn't guaranteed by the measurement board.

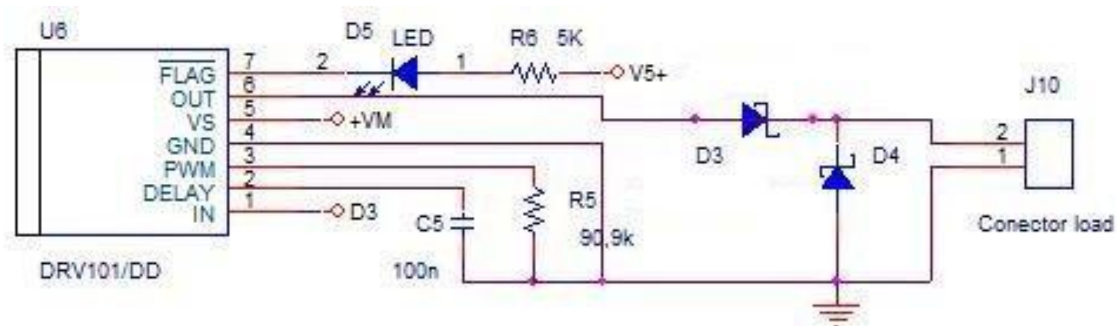


Figure 39: Schematic of DRV102

The layout was implemented using Ultiboard 10.1, distributed by National Instruments, and was thought considering the standard measures 10 cm x 10 cm, to apply our board at the measurement board downstream. Two layout's technology were used, the THT (Through Hole Technology) and the SMT (Surface Mount Technology), also called SMD.

These are the THT components used:

- 1 Trimmer of 500 kΩ;
- 1 Polarized capacitor of 22 µF;
- 2 Diodes 1N4007;
- 1 LED diode;
- 2 20Header connectors;
- 4 4Header connectors;
- 1 2Header connector;
- 1 6Header connector, adapted at 5 pins;
- 2 Banana connectors.

The SMT components used are:

- 4 instrumentation amplifiers AD620BRZ;
- 1 integrated voltage regulator TPS7101QDG4;
- 7 resistors of 1206 type;
- 6 capacitance 0f 0805 type;
- 1 driver DRV102.

Moreover on both of top and bottom sides was put a power plane, to guarantee the same connection to all the ground pins and to save some vias in the board.

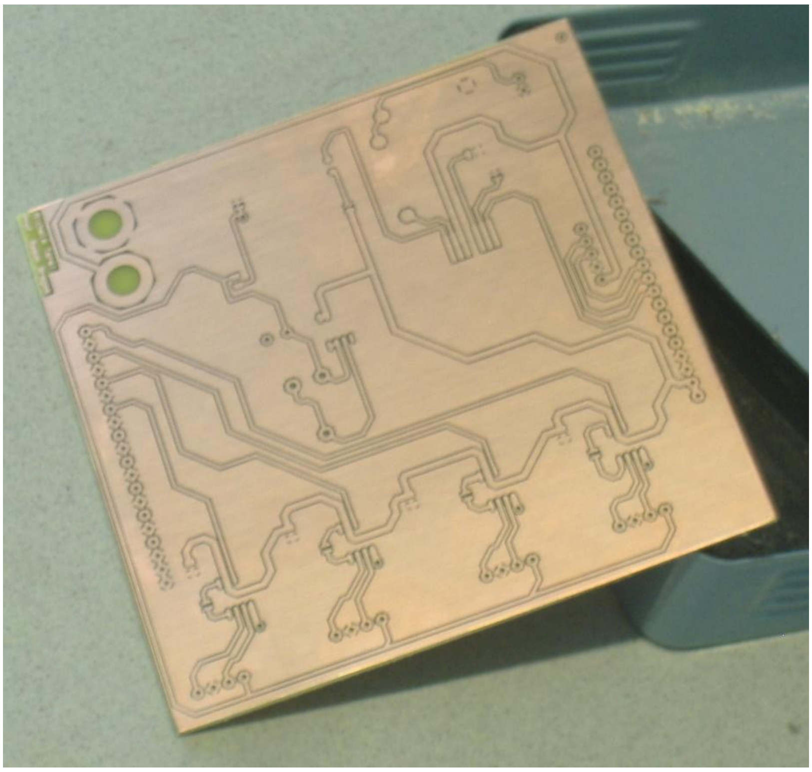


Figure 40: Copper Top of the circuit board

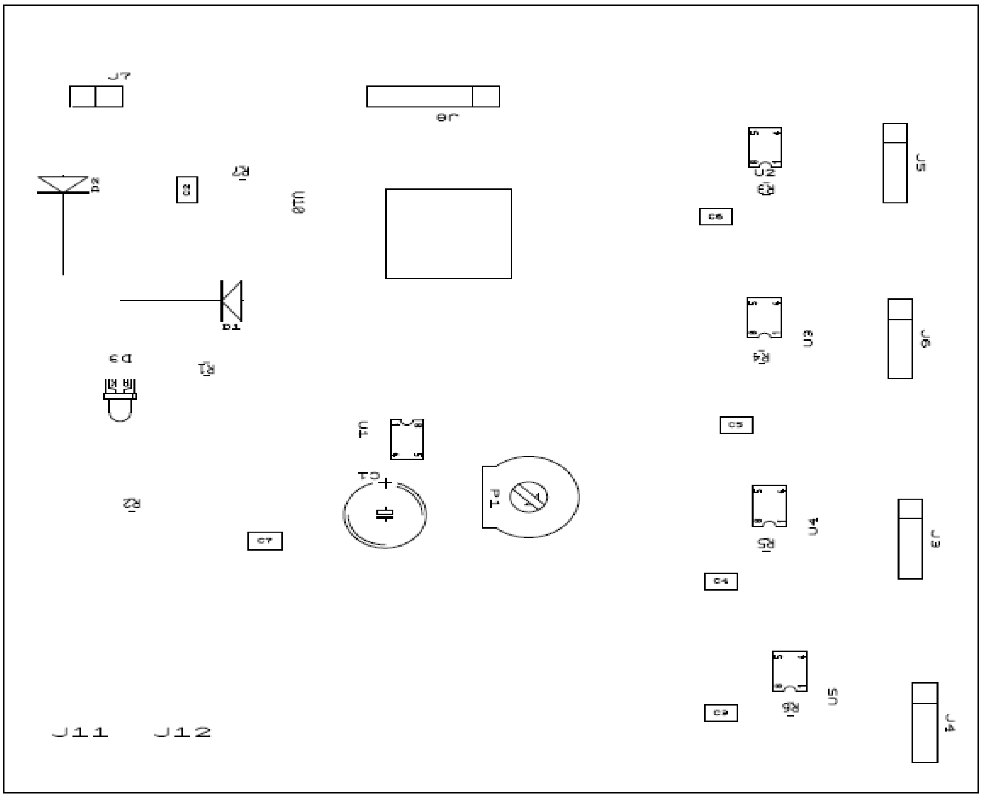


Figure 41: Footprint of the circuit board

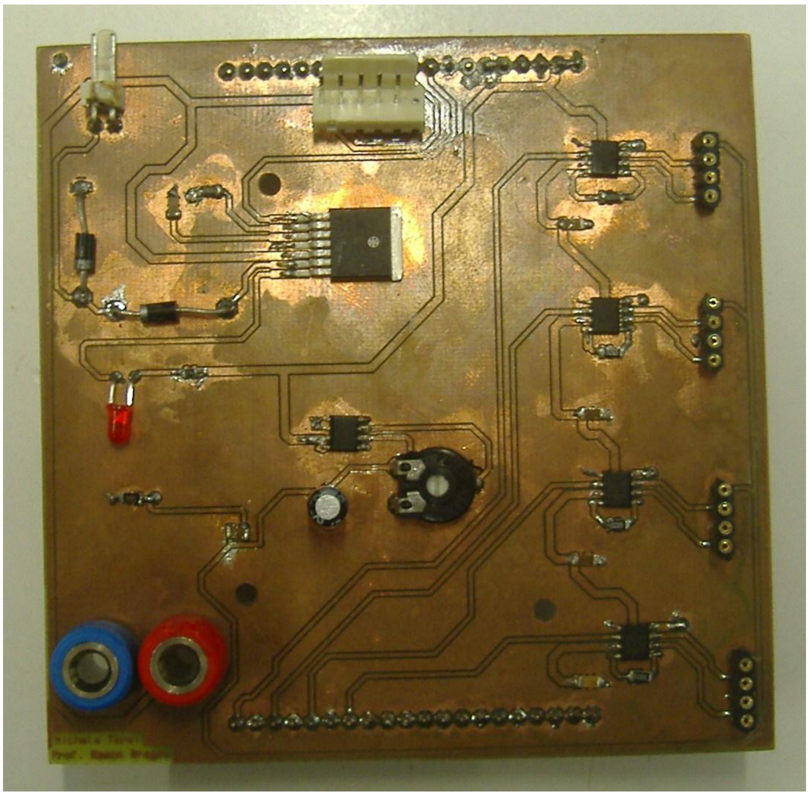


Figure 42: Physical realization of the circuit board

5.2 Control software

The control software was implemented using Labview 8.6, produced by National Instruments.
The main characteristic of this software is the initial division in static and dynamic measurements.

Static characterization

For this software was necessary to develop an internal driver to move the linear motor. To make this, the instructions already present in the Rabbit of measurement board were used.
The first step was to introduce a particular VI to set initially set all bits to zero, to start measurements in a stopped system.

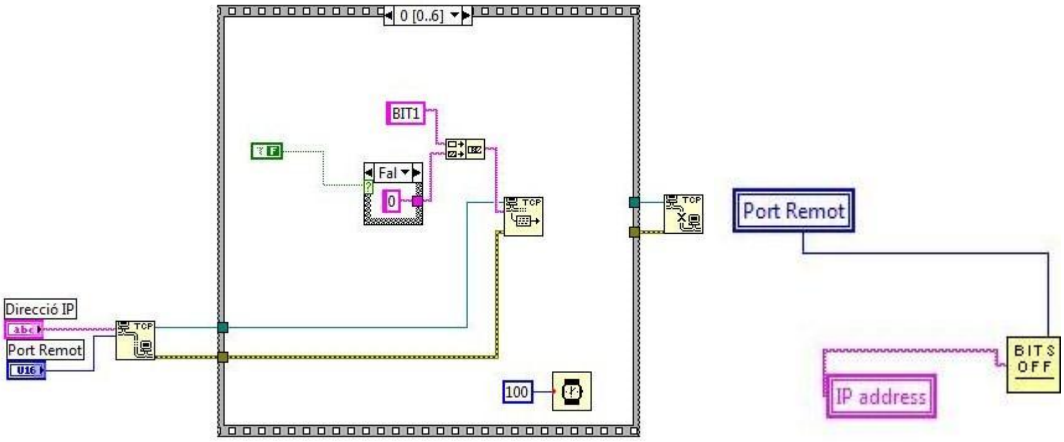


Figure 43: Bits Off software and how it appears in the final Labview measurement software

Then, using the Simple measure software, described in the 4th chapter, was implemented a little calibration part, to remove the initially off-set, maybe introduced during the setup of the catheter measurement system. This gives in output an indicator that can be used like a local variable, to subtract next this value to the final measure.

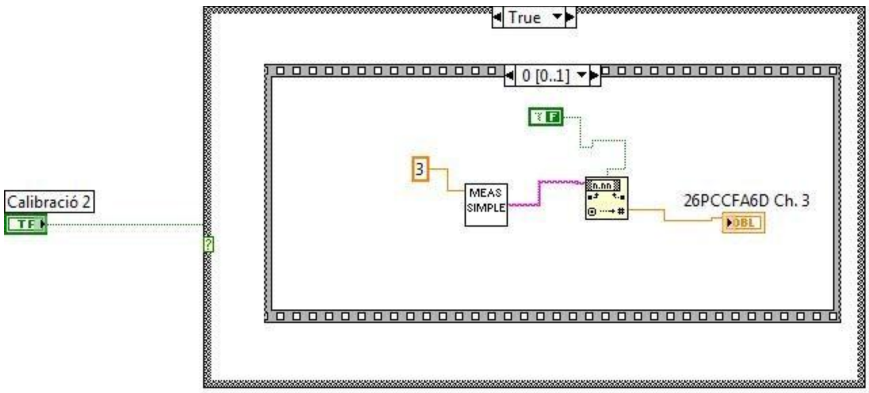


Figure 44: Calibration part

Then next step is the simple measure sequence, that use a single movement in a specified direction, and then makes the single measurement, considering the calibration variable, the gain of the instrumentation amplifiers and then the algebraic transformation to return in output a single value of the measured pressure. For moving the motor, with control buttons present in the VI, we can give also the length of the step to give more or less pressure.

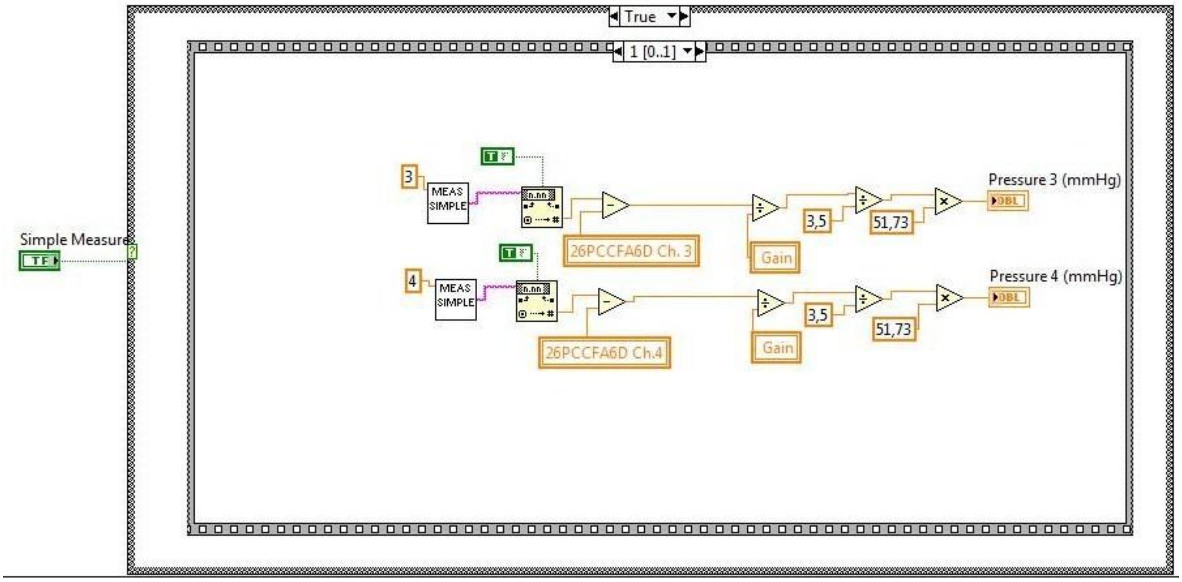


Figure 45: Simple Measure

The final step of this software is to create a part to move forward and backward the motor and the piston, always with the different length of steps option, to verify the real linearity of the system. To make this were used two for cycles, one for the forward moving, and one for the backward one. Each for cycle presents stacked sequence structures that give the possibility of saying before the direction and the length of the move, with the Move VI described before, and then of making the measurement of the pressure.

All the data measured were collected in a bundle, placed outside of the for cycles, to create a cluster with forward and backward measurement and then to draw an X-Y graph to verify the linearity of the system .

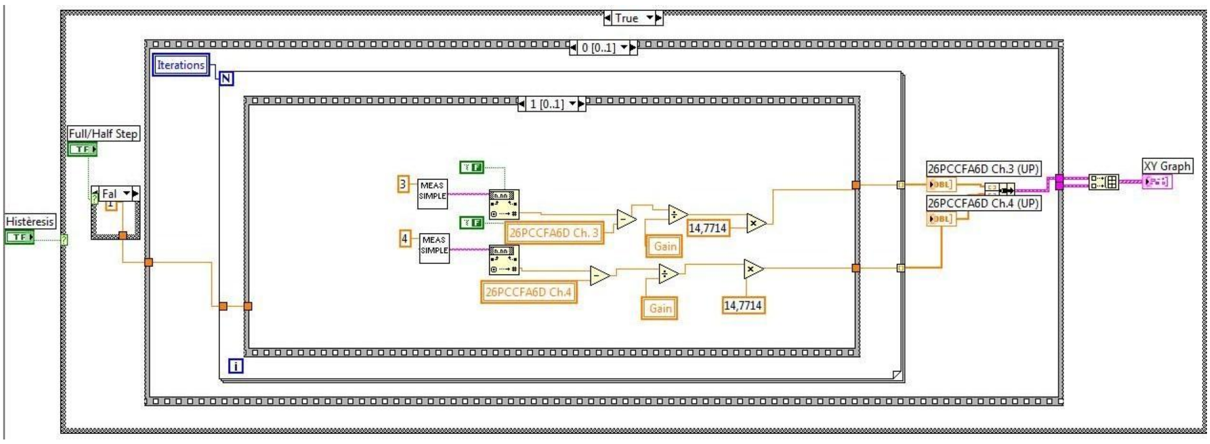


Figure 46: Moving UP part

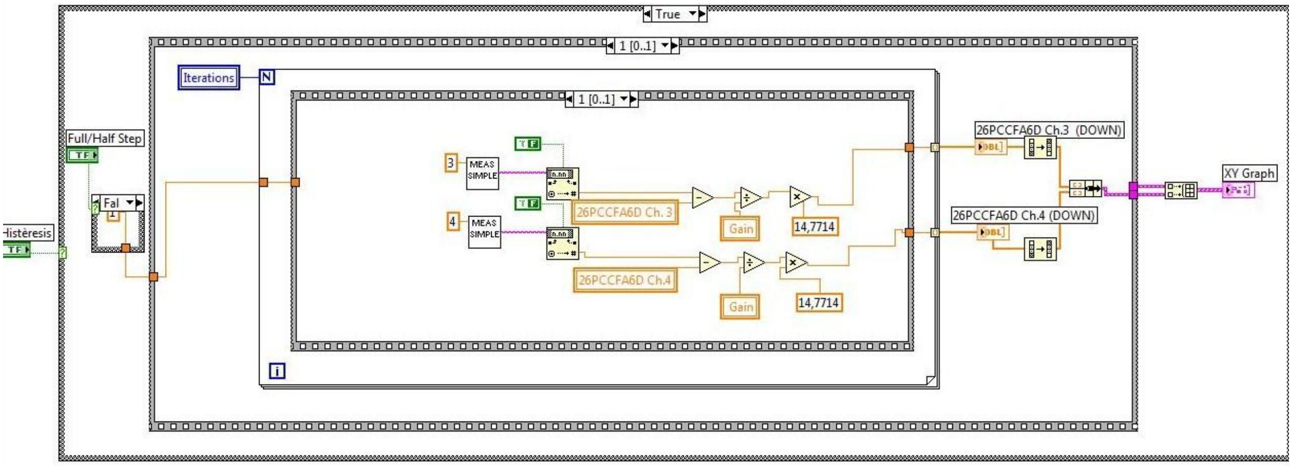


Figure 47: Moving down part

Dynamic characterization

Inside the same VI was also inserted the dynamic characterization, controllable like the static by a Boolean (True/False) controller.

This part starts, like the static one, with the Bits Off VI, to guarantee the initial conditions to the system, then also here we have the calibration system to remove the starting off-set.

Then, inserted in a while cycle, that repeat the measure until the stop, there is a flat sequence structure, to control the driver for the motor and to collect the measures.

The driver control is made by two little VI only to say to the Rabbit which bit we'll use to implement the clock, that we have to use and realize the clock, only putting the bit at 1 value or 0 value. In the middle of this sequences, there is a delay block, put also after the measurement VI, to control the clock period and to adapt this one to our requirements. Like shown in the figure, the Bit 4 (D3 in the measurement board) is used, using the command BITXY, where X is the number of the bit and Y is the value.

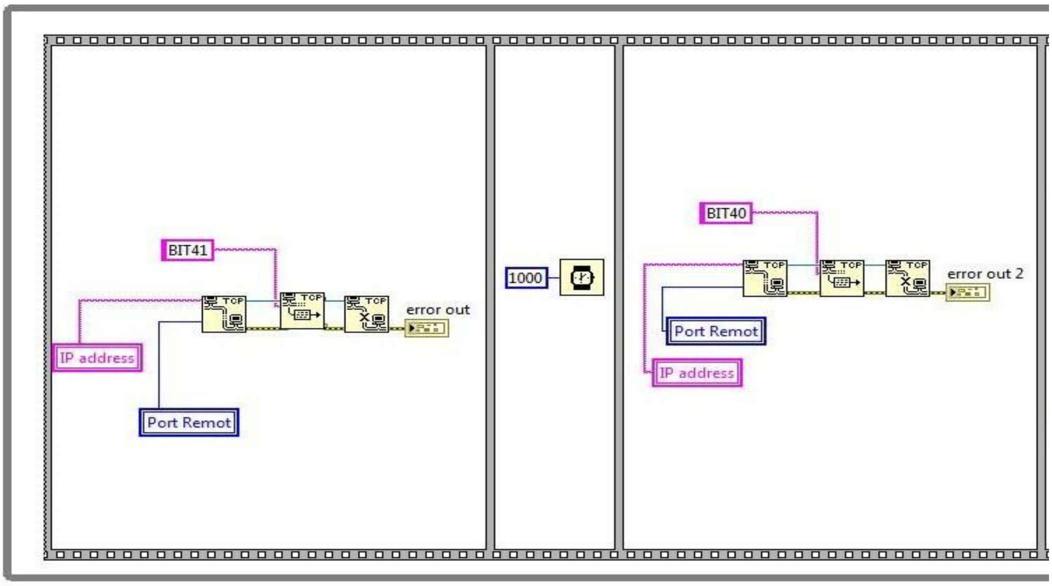


Figure 48: Bit controller for the driver

The next sequence regards the measurement VI, MeasureFastLV71, adapted on our demands, using only two channels, to leave the other two to the static measurement.

This VI is already described in the previous chapter, and is necessary to take fast measures, those measures will permit to us to take have a good characterization of the dynamic behavior.

The next sequence is another delay block, that is used to follow the clock already defined, giving the same delay of the previous one. To collect all the transitory, the FAST string part was put in the block, where is made the clock. This was due to the delay caused by the two TCP connections that was had to open. Now, during the clock signal goes down, the FAST block already begin to register the measures. Then it was possible to record good measures.

Moreover was introduced a block, that could save the vector produced by the measure VI. This could be useful for didactical purposes or for future offline signal processing with Matlab, for example.

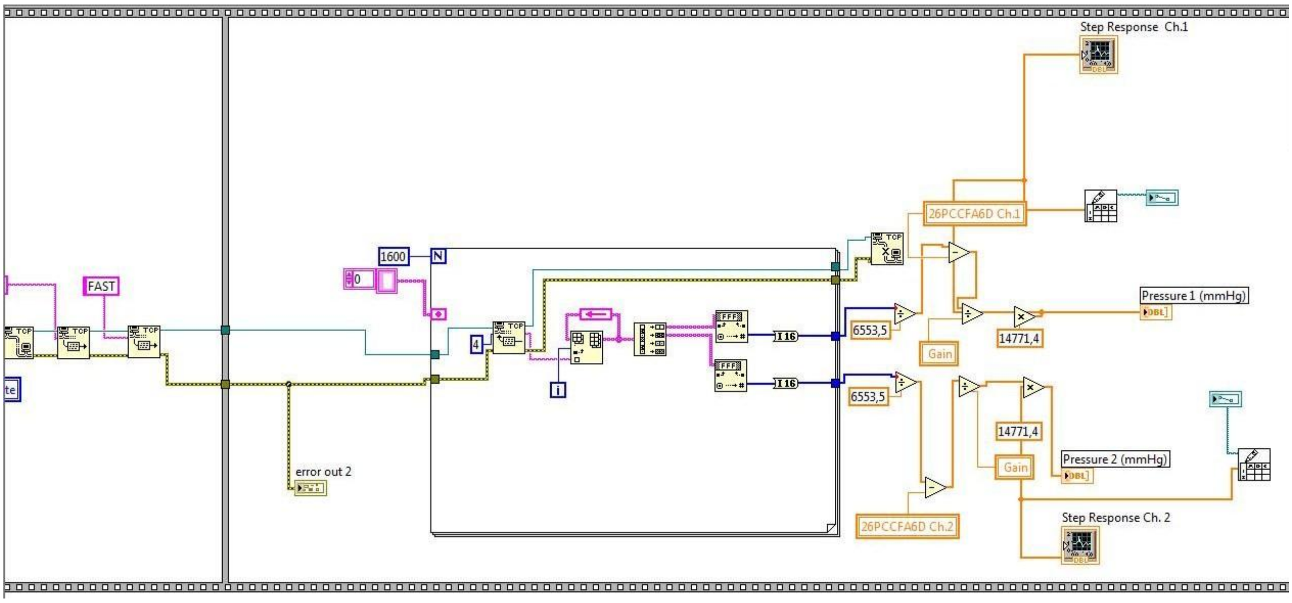


Figure 49: Measure fast VI

6. Characterization measurements

6.1 Measurements with air

The first step of our measurements were the static and dynamic measurements with catheters with air. Using the driver Labview to control our available tools, first it was obtained the static measurement, driving the microcontroller SMC CD85 to move the stepper motor. The goal is to demonstrate that, if the stimulus are slow enough, there is no other difference between the sensors located at both ends of the catheter that those derived from the static characteristics of the sensors, which are calibrable.

The next figure demonstrates how the system has a good linear static characteristic, with a linear regression: **output = 3,8 +0,9·input**.

This characteristic was taken moving the motor forwards and collecting the measures, then using the PE1 bit, available with the mixed-mode board, the motor was moved backwards and, like with the first movement the measures were collected in a bundle to create the ratio to insert in the XY graph.

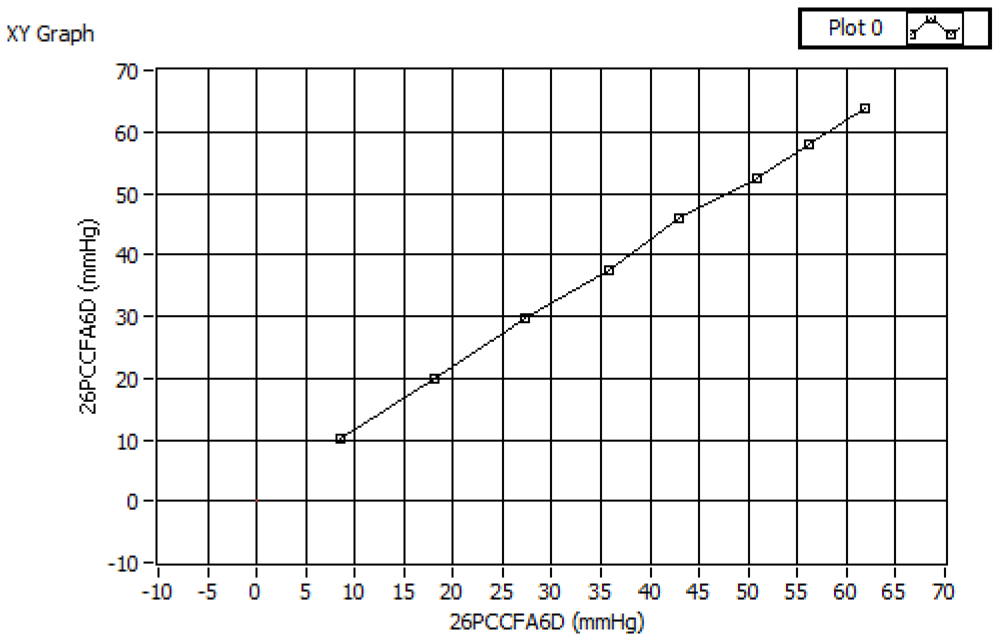


Figure 50: Static characteristic with air

Then were done the dynamic measurements, using the solenoid actuator.

Like with the testing board the second order response was very low, caused by the strong damping produced by the air.

Moreover was registered a noise at high frequency that disturbed the voltage signal, maybe due to the oscillation of the power supply, given to the sensors by the voltage regulatorTPS7101Q.

This problem was partially solved, exploiting the opportunity to save our resulting vector in a text file and using the Matlab function FILTFILT(B, A, X).

FILTFILT(B, A, X) filters the data in vector X with the filter described by vectors A and B to create the filtered data Y. The filter is described by the difference equation:

y(n) = b(1)*x(n) + b(2)*x(n-1) + ... + b(nb+1)*x(n-nb) - a(2)*y(n-1) - ... - a(na+1)*y(n-na)

After filtering in the forward direction, the filtered sequence is then reversed and run back through the filter; Y is the time reverse of the output of the second filtering operation. A double pass moving average filter can be obtained using this function.

The first image shows the dynamic response registered using a more large and elastic tube.

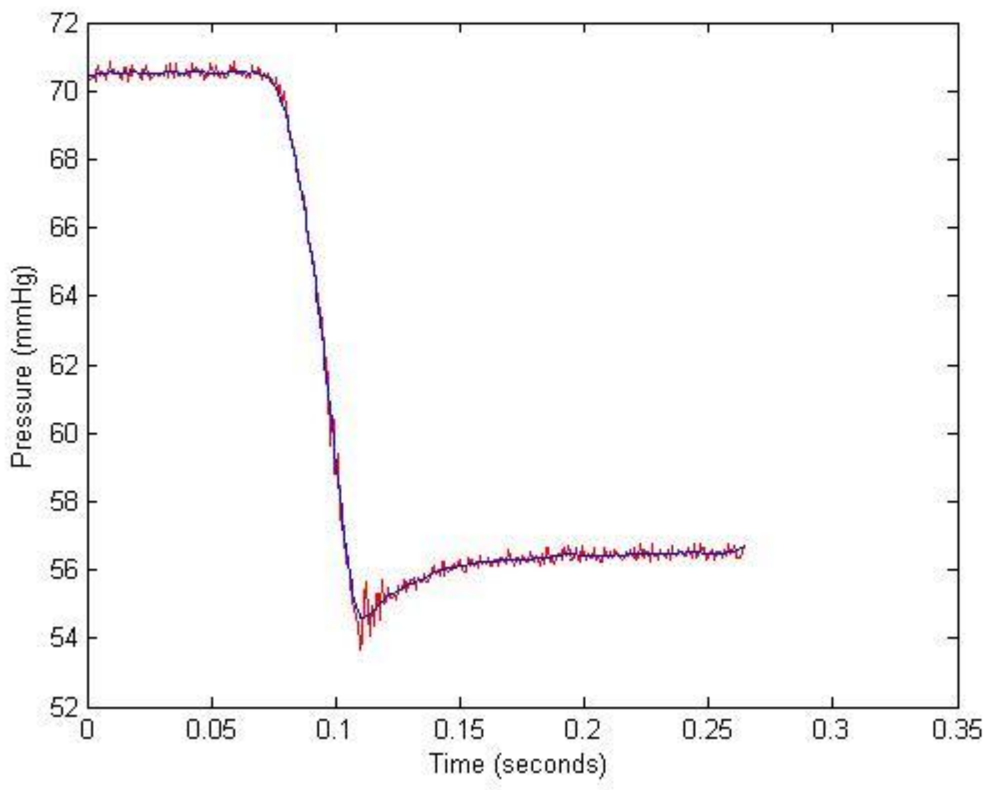


Figure 51: Filtered dynamic response with the elastic catheter

Then was made a measurement with a less large and elastic tube.

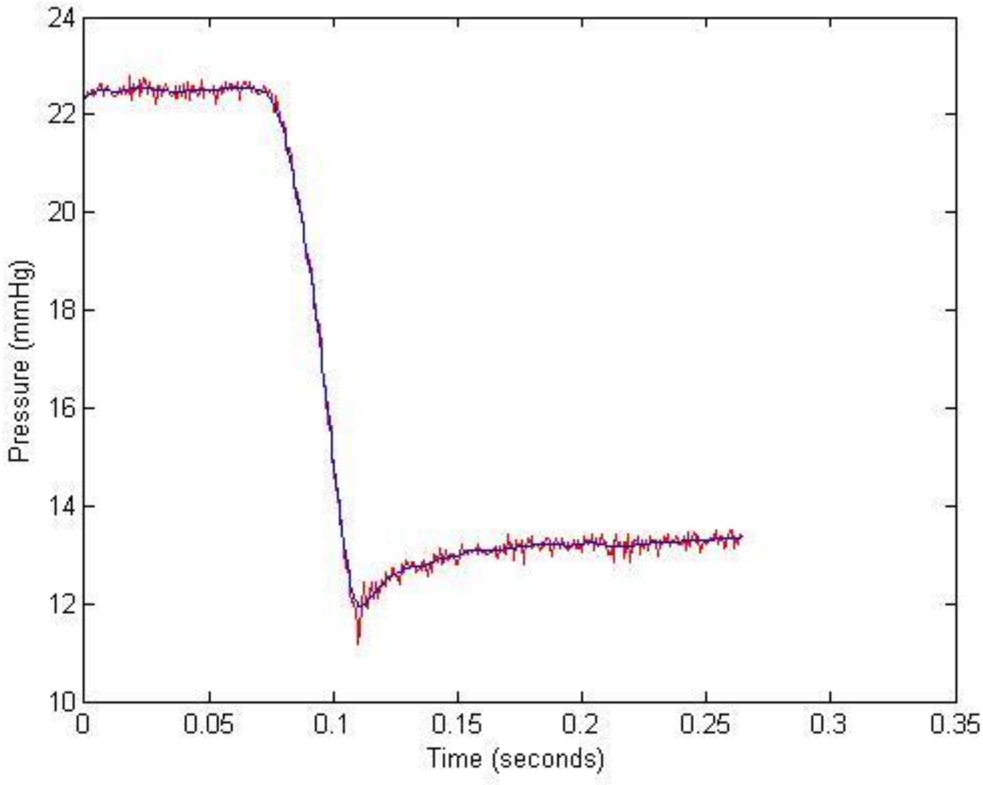


Figure 52: Filtered measurements with the more rigid catheter

The measurements show the ramp-like response of the solenoid actuator which partially hides the system response, followed by a exponential recovery with a time constant of approximately 25 ms. In any case, this is not a real situation given that the catheter and sensors are designed to work with liquids.

6.3 Measurement with liquid and liquid+ air bubbles

The next step was to collect measurements using a liquid, in this case the water, and adding air bubbles to emulate the effect of a bubble captured in the catheter.

It was first obtained the static measurement with the stepper motor. Inside of the piston was put water. Here was collected a linear characteristic, to view if the static behavior was better respect to the system with only air. The next figure shows the XY graph of these measurements. The linear regression gives a characteristic **output = 117,25 + 0,95 · input**, with a zero error and sensitivity which are calibrable.

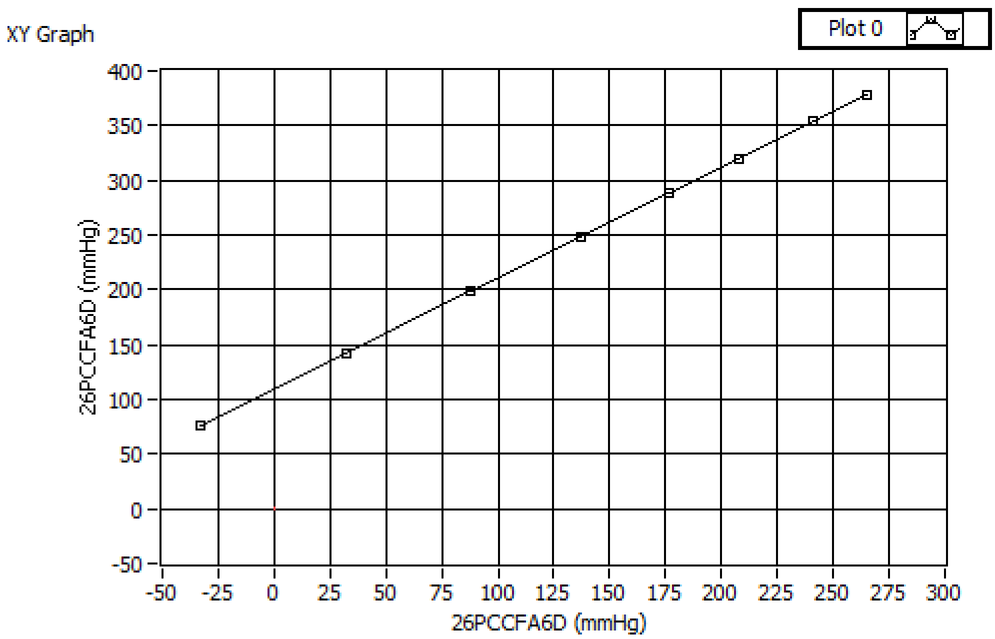


Figure 53: Static characteristic with water

For the dynamic measurements was still used the inductive motor. In one tube were put air bubbles and in the other only water.

To measure the behavior with air bubbles were used the first channel, available in the measurement board. The more large and elastic catheter was the ideal to realize the system with air and finally was registered a first good dynamic response for our system. Like could be seen in the next figure, due to the air bubbles, the second order response is more damped and presents a period of 95,7 ms, frequency of 12,1 Hz and a damping factor of 0,091.

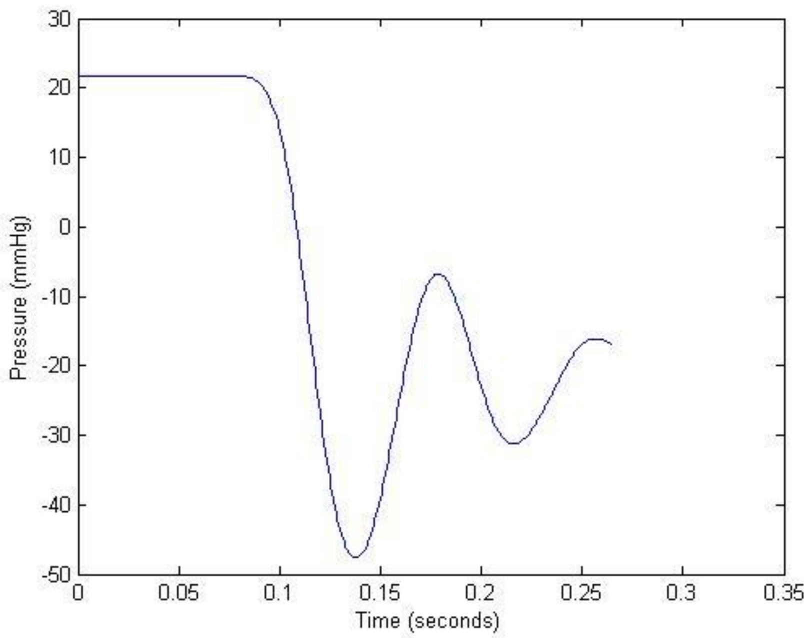
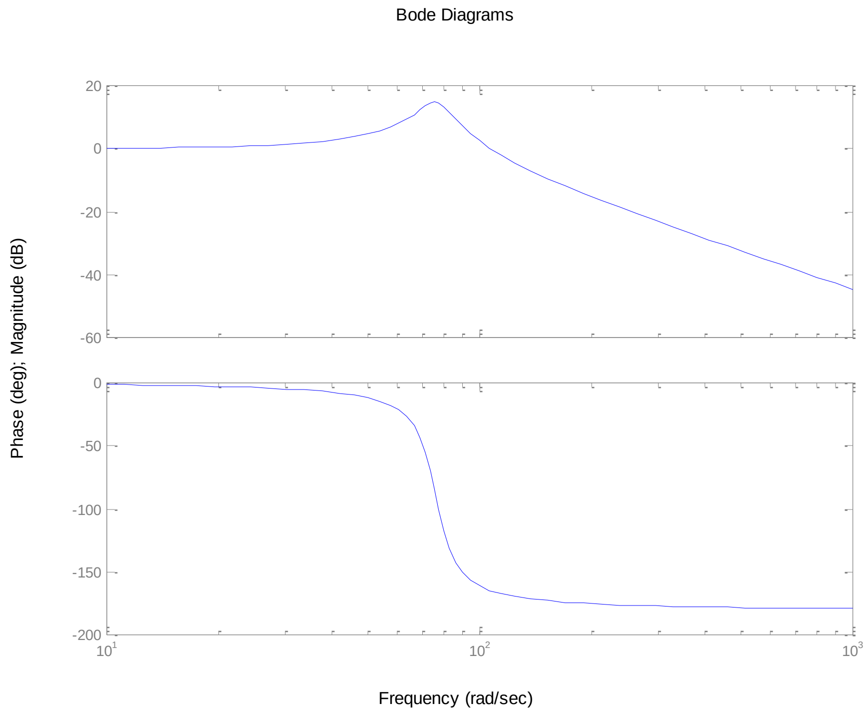


Figure 54: Filtered dynamic response for catheter with bubble of air

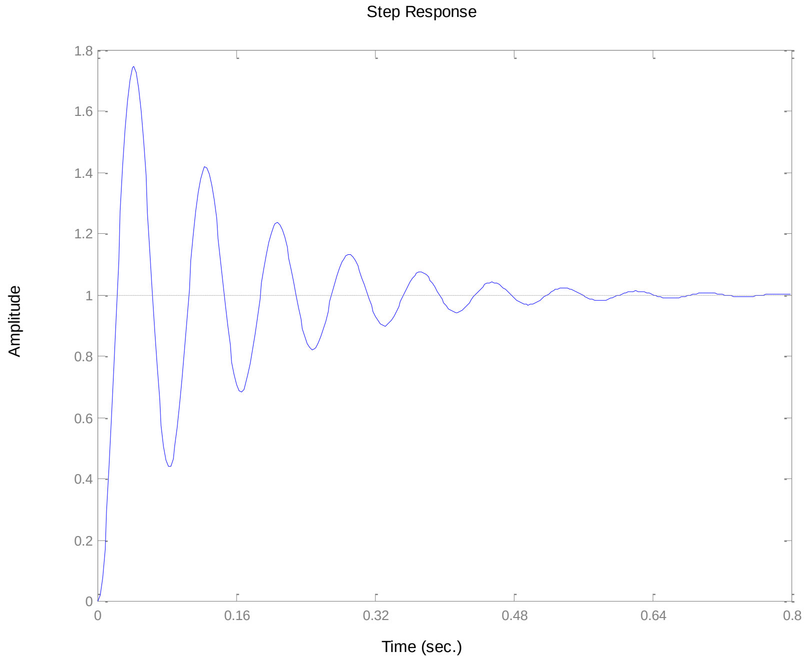
The system identification using the method described in chapter 2, gives a second order response with parameters ξ= 0.0912 and $f_0 =$ 12.16 Hz. This provides a transfer function

$$H(s) = \frac{5828}{s^2 + 13{,}92\ s + 5828}$$

With a Bode representation:



And a step response:



The channel two was used for the only water measurement.

To collect the measurement with only water was used the rigid and less large catheter. In this measurement was registered a less damped response with a period of 58,65 ms, a frequency of 17,1 Hz and a damping factor of 0,055. Then we can say that with only water there a less damped step response. The step response looks good, respect of the theoretical one.

The time division is made considering a sample frequency of 1,5 kHz

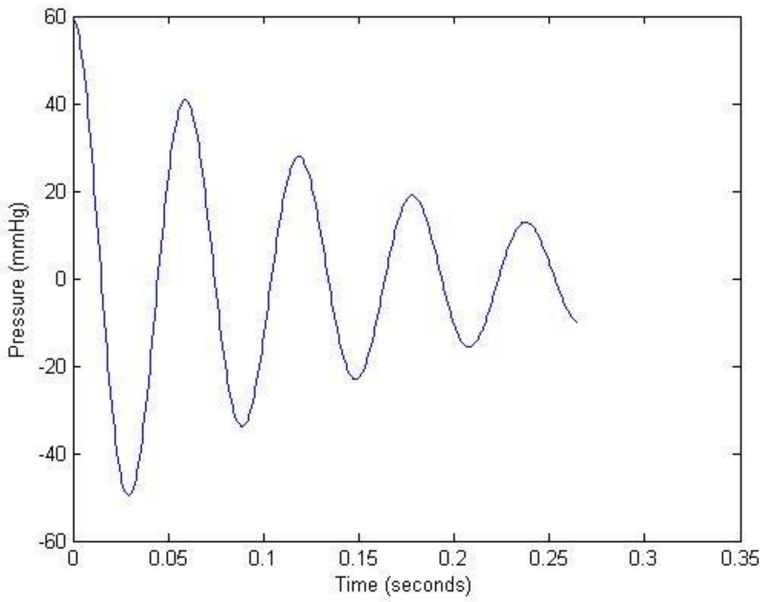
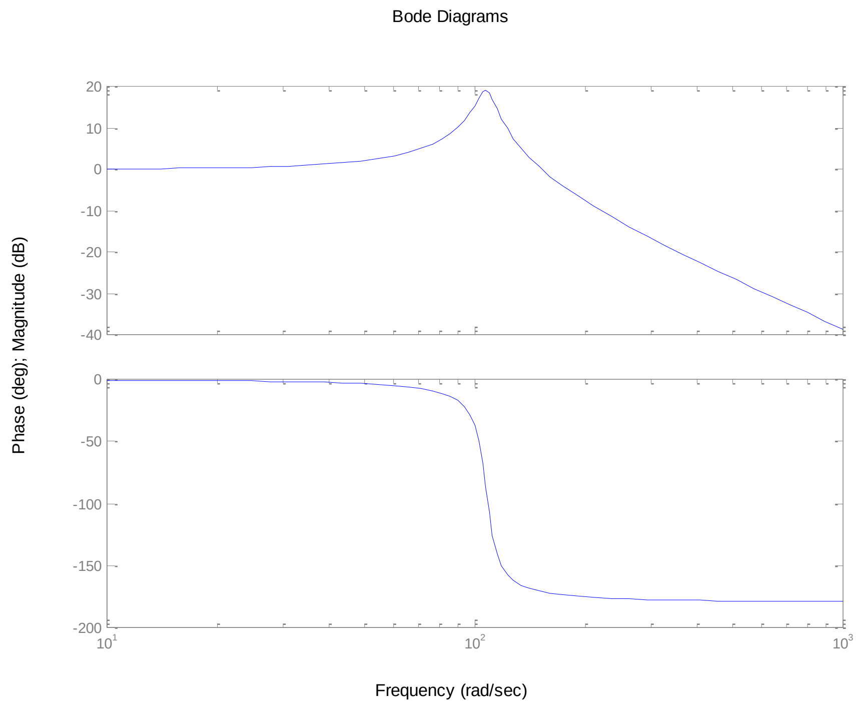


Figure 55: Filtered Dynamic response with only water

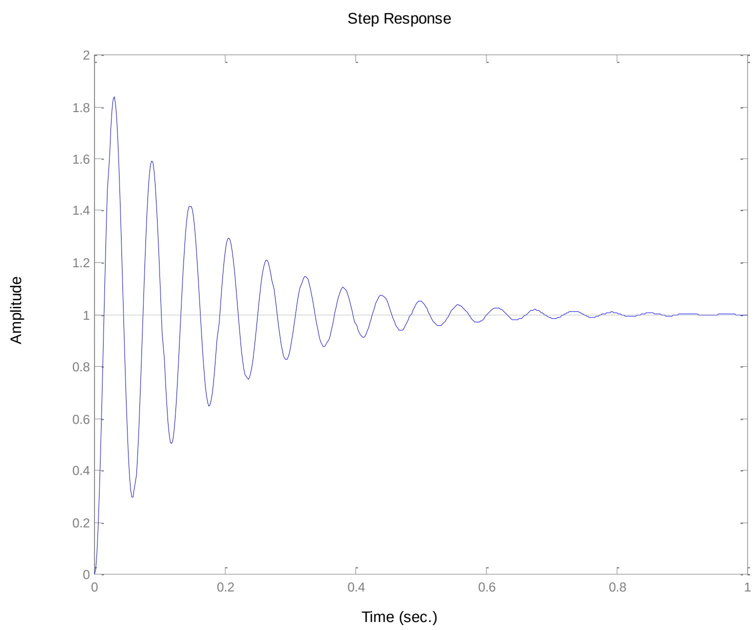
The system identification using the method described in chapter 2, gives a second order response with parameters ξ= 0. 0.055 and $f_0 =$ 17.08 Hz. This provides a transfer function

$$H(s) = \frac{1{,}15e4}{s^2 + 11{,}89\ s + 1{,}15e4}$$

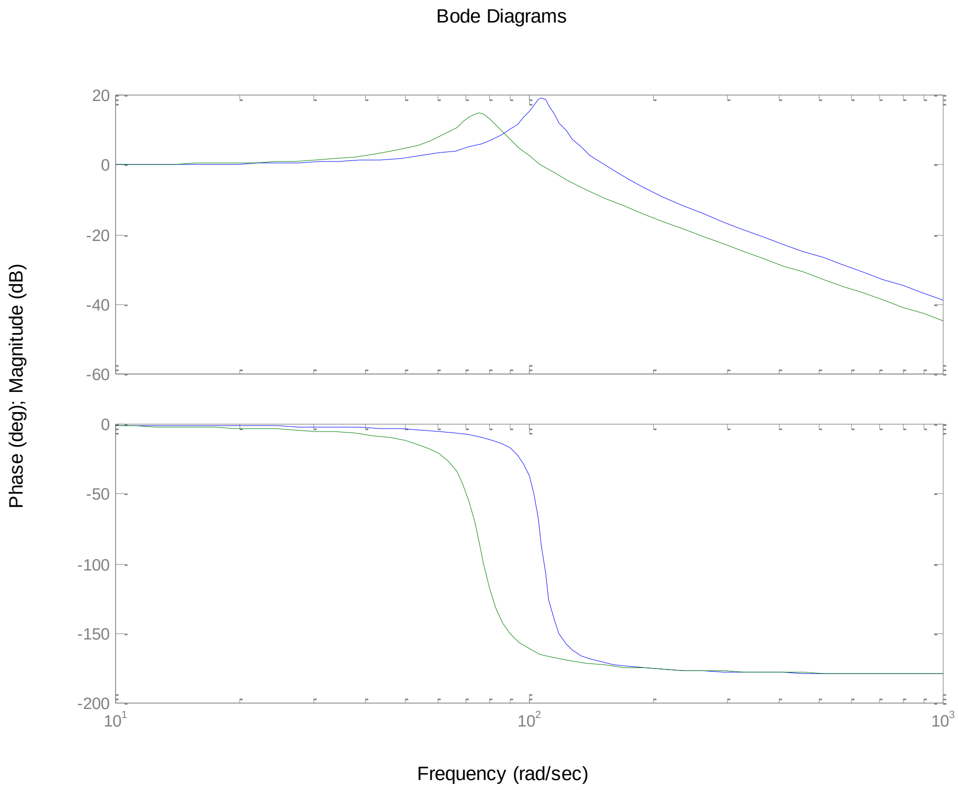
With a Bode representation:



And a step response:



The joint representation of both frequency responses allow to compare them and display the effect of bubbles, giving a response similar to that shown in chapter 2 (figure 16):



7. Conclusions and future developments

In this project was developed a system for the characterization of the static and dynamic response of a catheter-sensor system. This idea derives from the study of the invasive blood pressure measurements, analyzed in the Instrumentation and Biomedical Engineering of the Electronics Department of the UPC in Barcelona. Using an inductive motor (solenoid) was reproduced an impulse to excite the catheter-sensor system, once with air filling and once with water filling. Moreover with water in the catheters, it was analyzed the behavior with bubbles of air in the system.

The results were like was expected, with a strong damped response when in the catheters were only air, with a very low resonant frequency.

Using water, that don't damp the signal like the air, the results were like we expected, measuring a damping factor and a period higher with air bubbles in water than with water only.

Moreover, using the mixed-mode board, this project will be useful for didactical purposes, profiting of the Ethernet connection of this board with the servers.

For the future it will be good to substitute the plastic tube with real catheters, to improve the catheter-sensor system and to adapt this to the real blood pressure measurement, or to a real testing system for the catheters. Besides it's better to use more large catheters to avoid some signal rebounds that could sum the oscillations and give a disturbed response.

Moreover it will be useful to substitute the water with a better liquid, like already said a good candidate could be "Oraldine", that won't degenerate with the time nor contaminate the system.

Given that the purpose of the system is didactical, large improvements can be done by using more sophisticated system identification techniques.

8. Bibliography

[1] “Medical Instrumentation, Application and Design” John G. Webster, Ed. John Wiley, 1998 (3ª ed).

[2] http://www.medphys.ucl.ac.uk/teaching/undergrad/projects/2003/group_03/history.html

[3] “The Biomedical Engineering Handbook”, Joseph D. Bronzino, Editor-In-Chief 2000 (2ª ed).

[4] “Medical Physics Biomedical Engineering”, B.H. Brown, R.H Smllwood, D.C. Barrer, P.V. Lawford, D.R. Hose, Ed. Institute of Physics, 1999.

[5] “Modular Workbench for In-Situ and Remote Laboratories”, Benjamin Sanchez and Ramon Bragos, Technical University of Catalonia (UPC). ETSETB. Electronic Engineering Department.

[6] Sanchez Terrones, Benjamin (2006). Projecte Final de Carrera: Sistema modular per la realització de pràctiques presencials i remotes en entorns d’instrumentació. Barcelona:ETSETB

9. Attachments

9.1 Orcad schematic

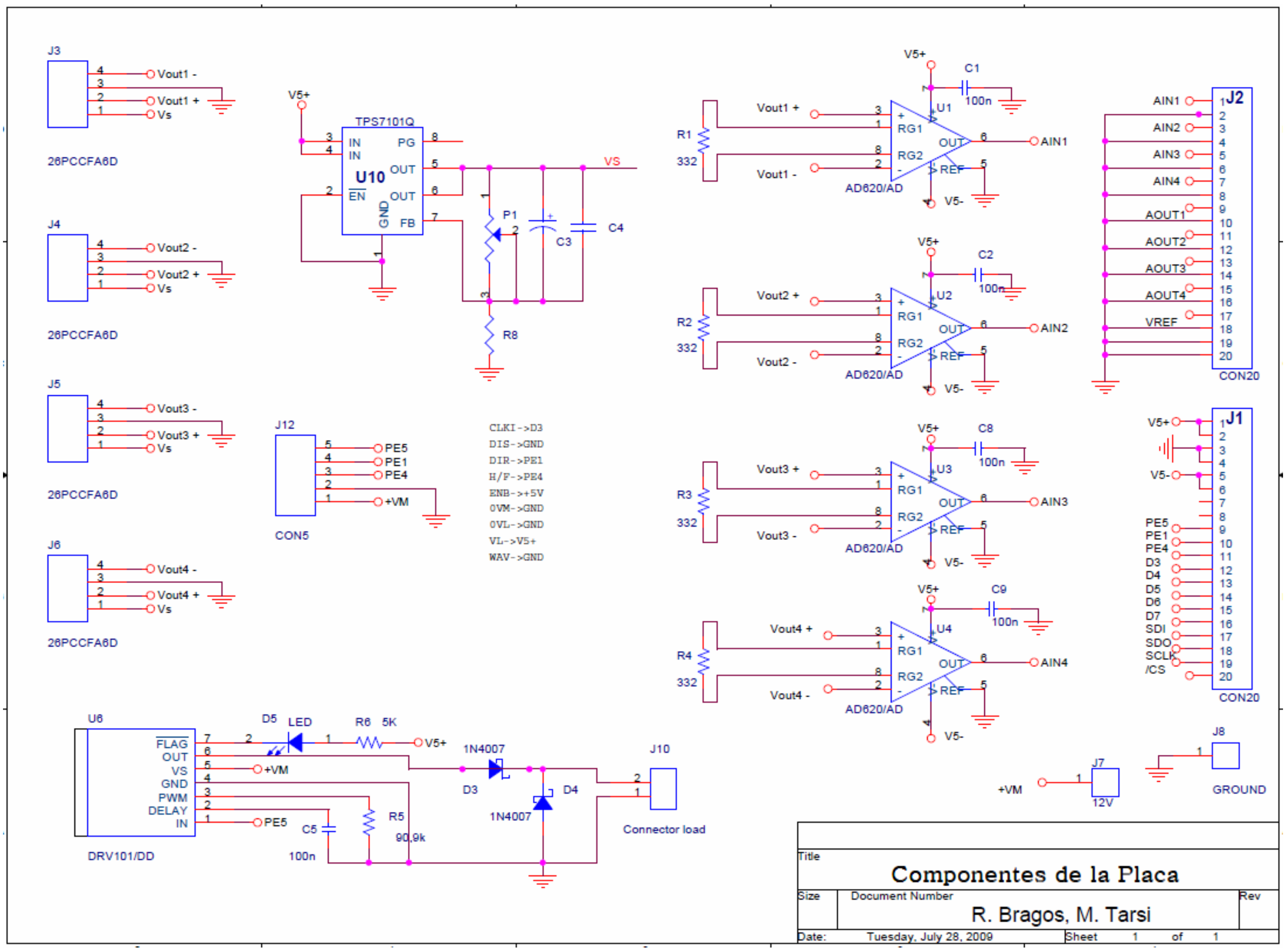


Figure 56: Orcad schematic of the circuit board

9.2 Layout

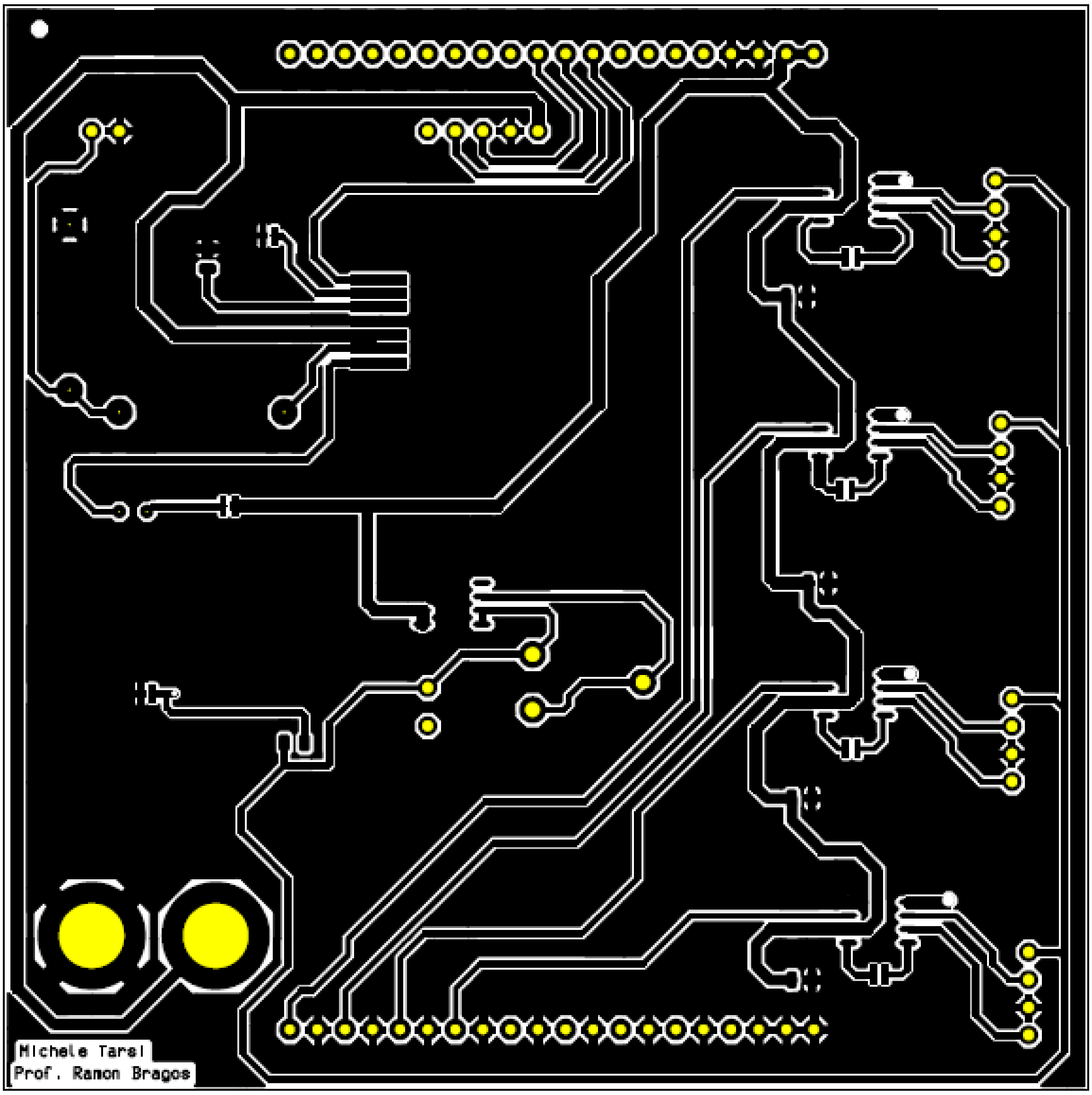


Figure 57: Copper Top of the circuit board

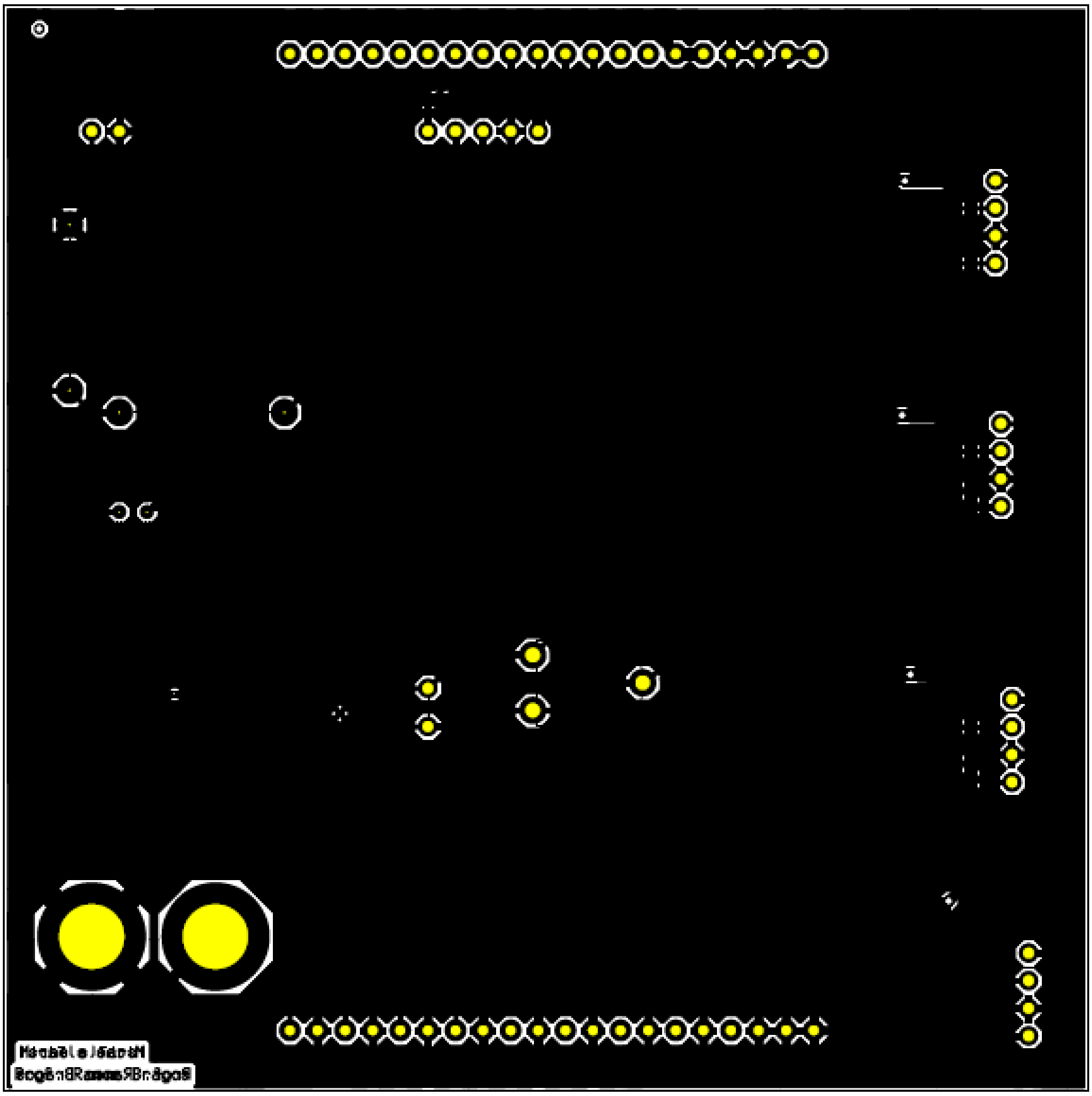


Figure 58: Copper Bottom of the circuit board

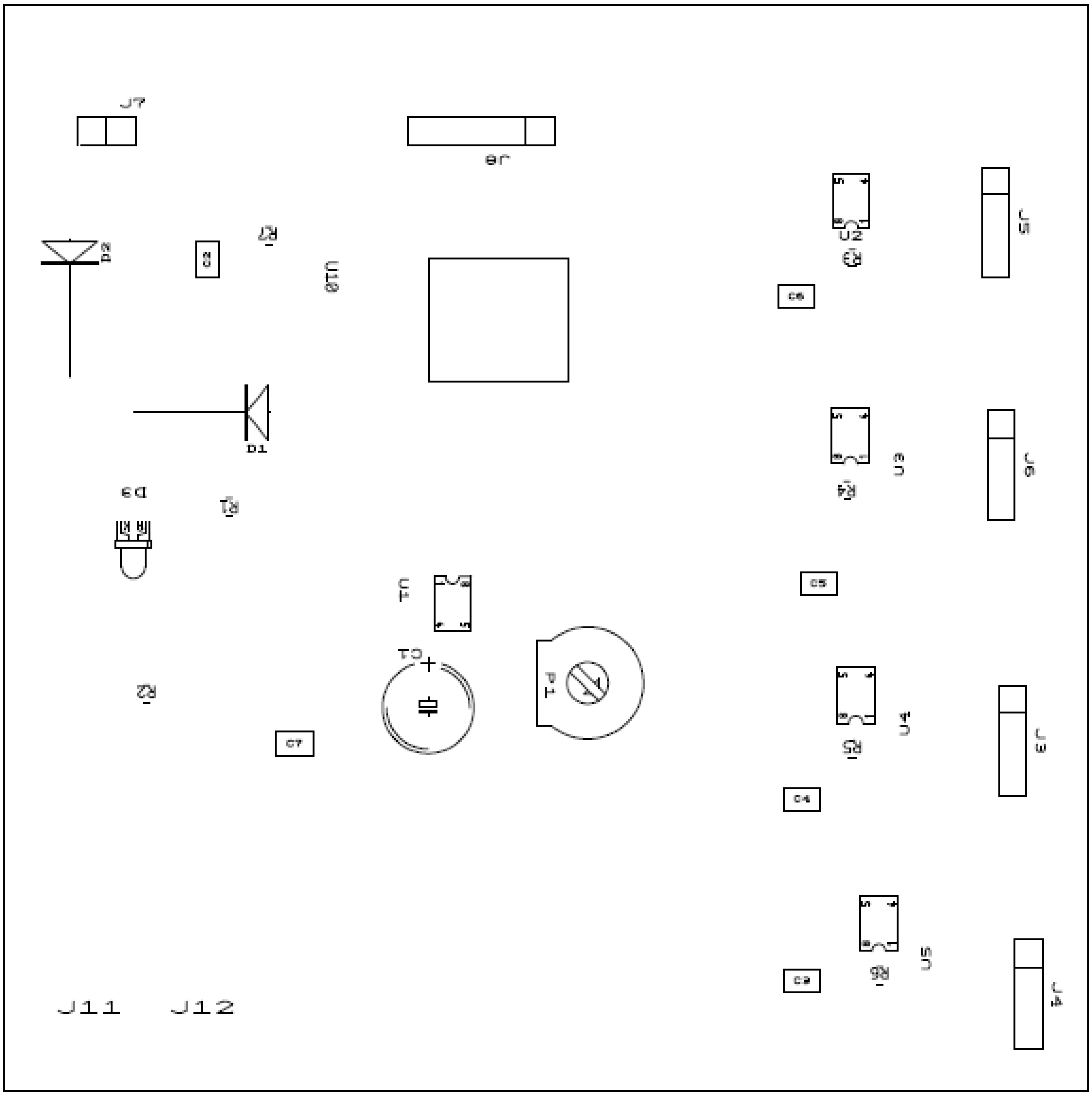


Figure 59: Footprint of the circuit board

9.3 Components' list

Quantity	Part Reference	Value
1	C1	100 nF
1	C2	100 nF
1	C3	Cap Pol
1	C4	Cap NP
1	C5	100 nF
1	C8	100 nF
1	C9	100 nF
1	J1	CON20
1	J2	CON20
1	J3	26PCCFA6D
1	J4	26PCCFA6D
1	J5	26PCCFA6D
1	J6	26PCCFA6D
1	J7	12 V
1	J8	GROUND
1	J10	Connector Load
1	J12	CON5
1	P1	RESISTOR VAR
1	R1	332 Ω
1	R2	332 Ω
1	R3	332 Ω
1	R4	332 Ω
1	R5	90.9 kΩ
1	R6	5 kΩ
1	R8	560 kΩ
1	U1	AD620
1	U2	AD620

1	U3	AD620
1	U4	AD620
1	U6	DRV102
1	U10	TPS7101Q
1	D3	1N4007
1	D4	1N4007
1	D5	LED

9.4 Driver Labview

Controls and indicators

IP Address

Port Remote

Iterations

26PCCFA6D Ch.1

26PCCFA6D Ch.2

26PCCFA6D Ch.3

26PCCFA6D Ch.4

Pressure 1 (mmHg)

Pressure 2 (mmHg)

Pressure 3 (mmHg)

Pressure 4 (mmHg)

Gain

Steps

ON

Static Measure

Dynamic measure

Calibration1

Calibration2

Stop

Go/Back

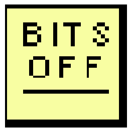
Full/Half Step

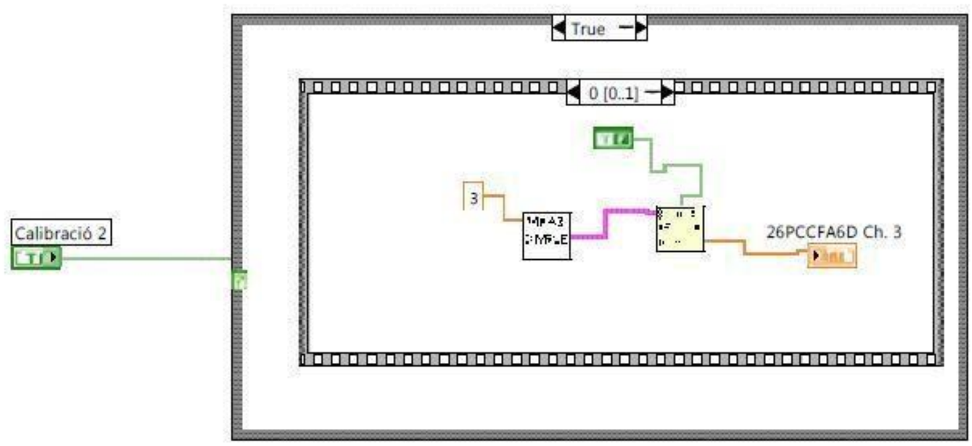
Hysteresis

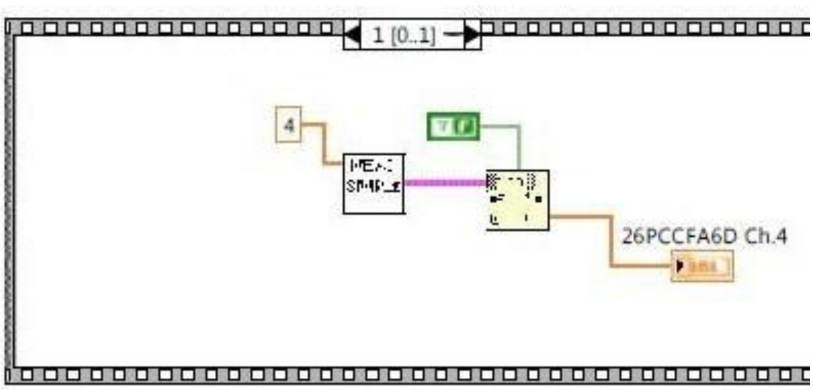
Simple Measure

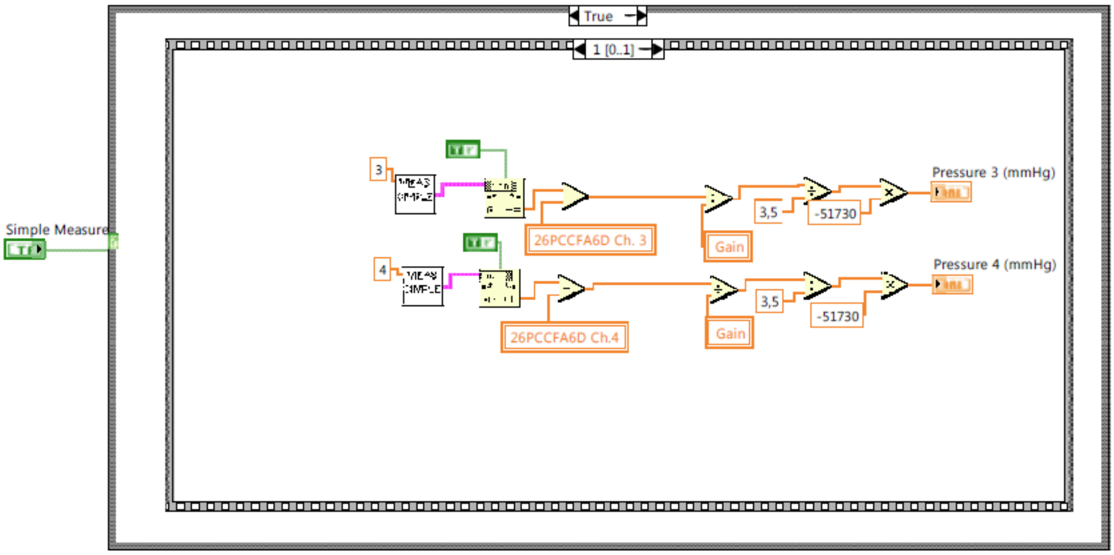

XY Graph

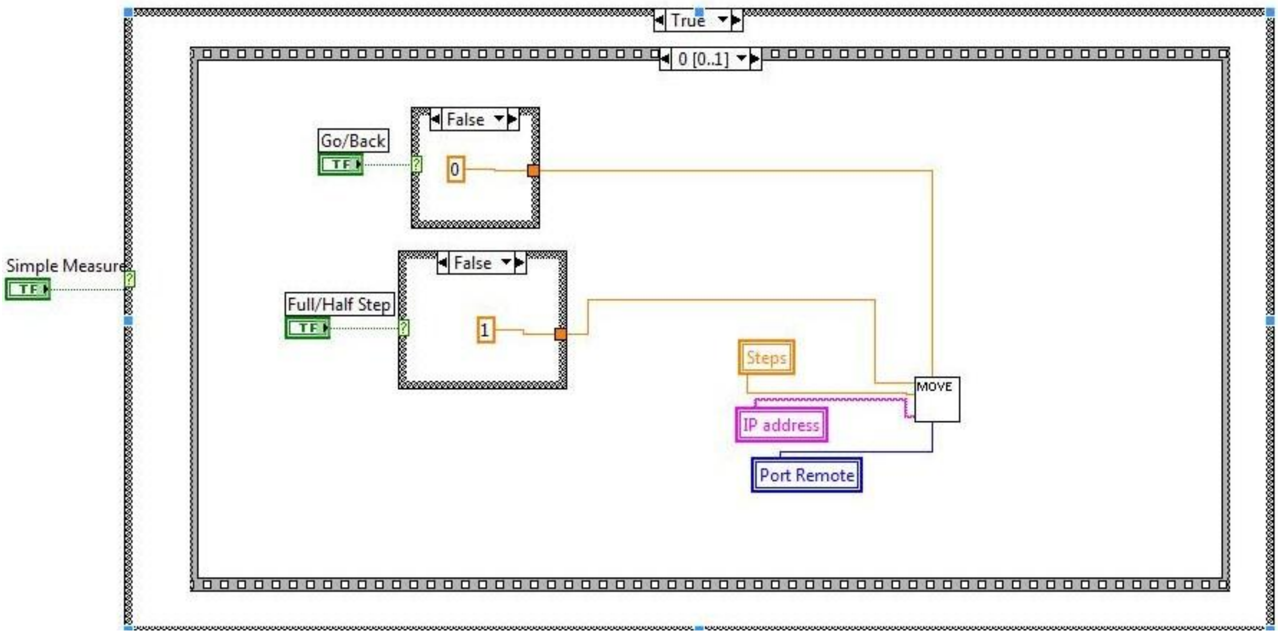
Block Diagram

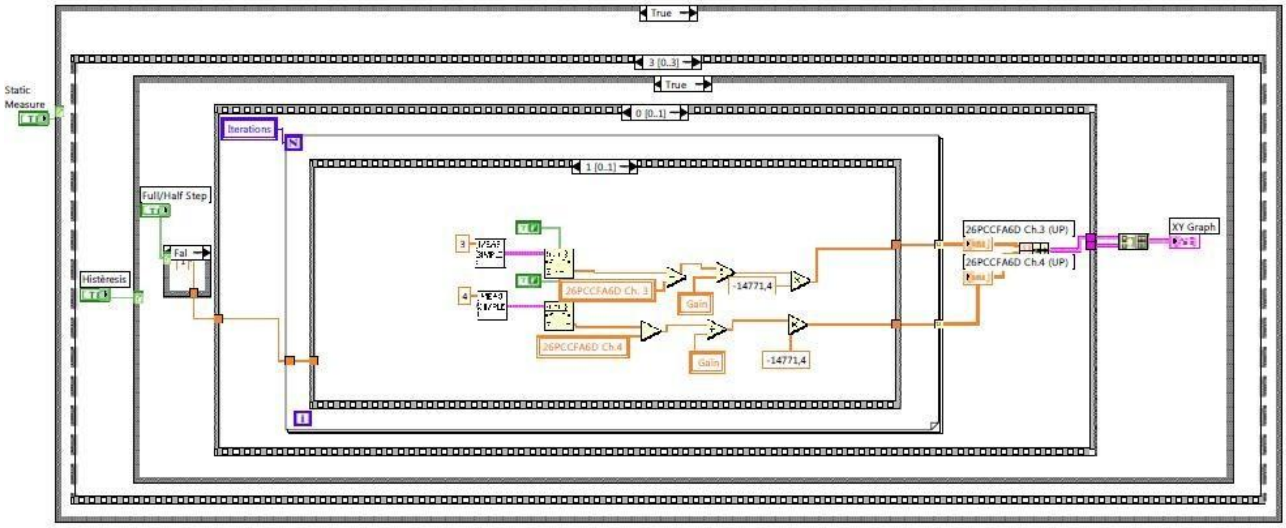


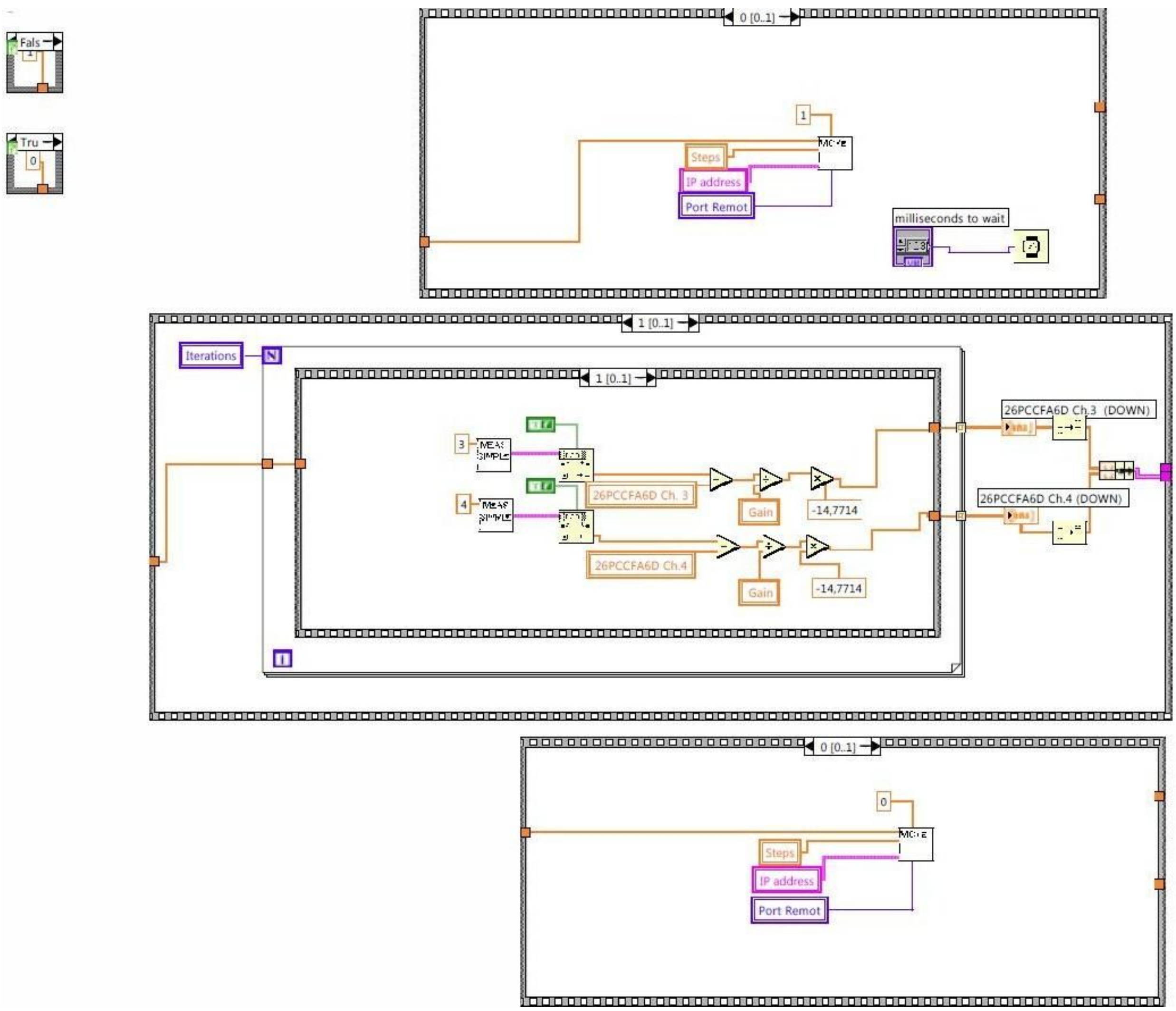


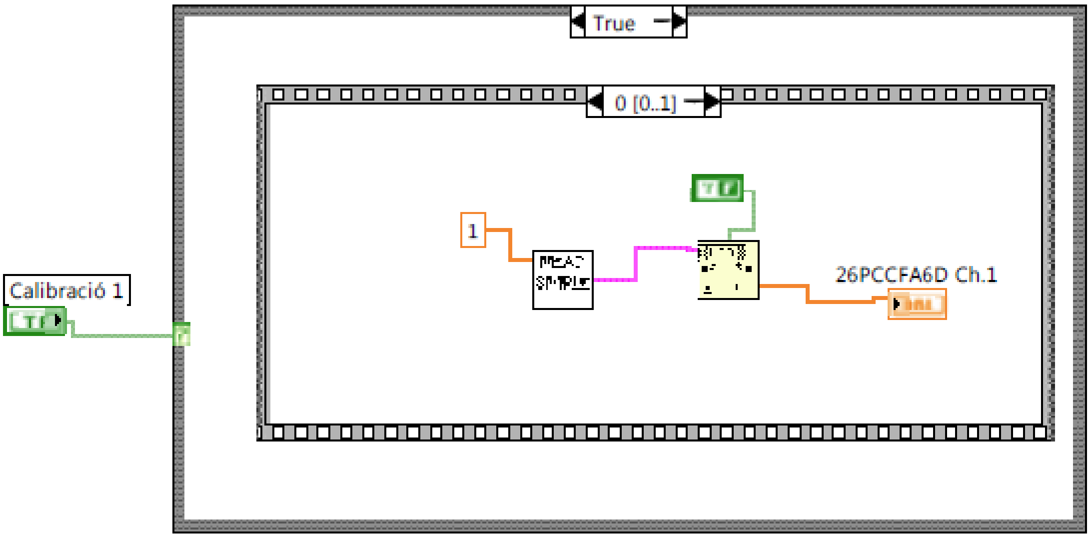


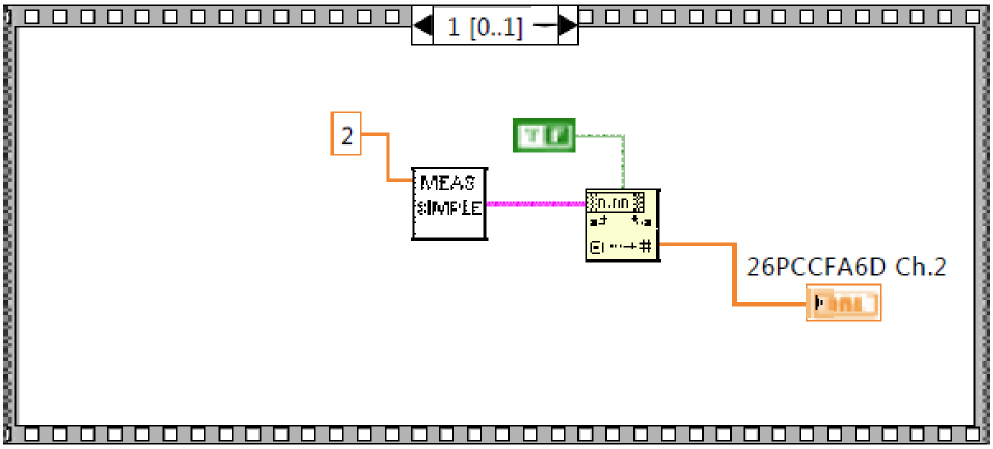

True
1 [0..1]
Simple Measure
3
4
26PCCFA6D Ch. 3
26PCCFA6D Ch.4
Gain
Gain
3,5
3,5
-51730
-51730
Pressure 3 (mmHg)
Pressure 4 (mmHg)

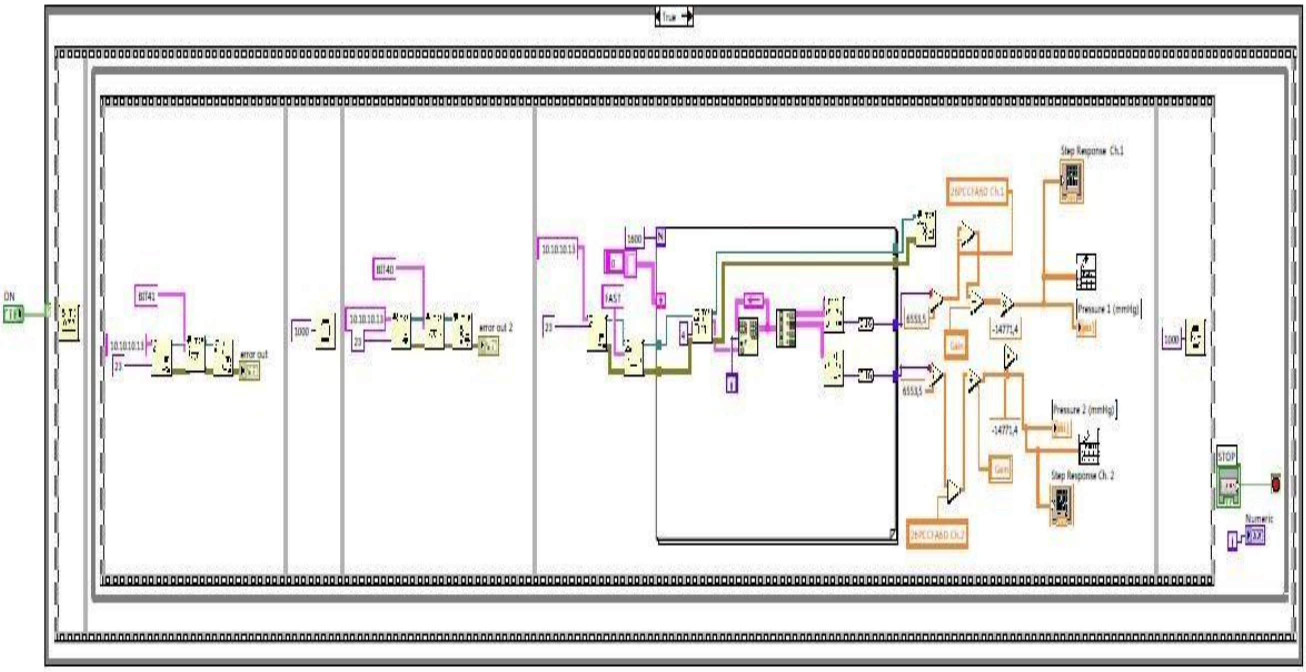

True
0 [0..1]
Go/Back
False
0
Simple Measure
Full/Half Step
False
1
Steps
MOVE
IP address
Port Remote


True
3 [0..3]
True
0 [0..1]
Static Measure
Iterations
1 [0..1]
Full/Half Step
Histèresis
26PCCFA6D Ch. 3
26PCCFA6D Ch.4
Gain
-14771,4
26PCCFA6D Ch.3 (UP)
26PCCFA6D Ch.4 (UP)
XY Graph


0 [0..1]
Steps
IP address
Port Remot
milliseconds to wait
1 [0..1]
Iterations
26PCCFA6D Ch. 3
26PCCFA6D Ch.4
Gain
-14,7714
26PCCFA6D Ch.3 (DOWN)
26PCCFA6D Ch.4 (DOWN)
0 [0..1]
Steps
IP address
Port Remot


True
0 [0..1]
1
MEAS SIMPLE
26PCCFA6D Ch.1
Calibració 1


1 [0..1]
2
MEAS SIMPLE
26PCCFA6D Ch.2


True
ON
BIT41
error out
1000
BIT40
error out 2
FAST
1600
Step Response Ch.1
Pressure 1 (mmHg)
Pressure 2 (mmHg)
Step Response Ch. 2
Gain
-14771,4
STOP
Numeric

9.5 Components' specifications

AD620–SPECIFICATIONS (Typical @ +25°C, V_S = ±15 V, and R_L = 2 kΩ, unless otherwise noted)

Model	Conditions	AD620A Min	AD620A Typ	AD620A Max	AD620B Min	AD620B Typ	AD620B Max	AD620S[1] Min	AD620S[1] Typ	AD620S[1] Max	Units
GAIN	G = 1 + (49.4 k/R_G)										
Gain Range		1		10,000	1		10,000	1		10,000	
Gain Error[2]	V_{OUT} = ±10 V										
G = 1			0.03	0.10		0.01	0.02		0.03	0.10	%
G = 10			0.15	0.30		0.10	0.15		0.15	0.30	%
G = 100			0.15	0.30		0.10	0.15		0.15	0.30	%
G = 1000			0.40	0.70		0.35	0.50		0.40	0.70	%
Nonlinearity,	V_{OUT} = –10 V to +10 V,										
G = 1–1000	R_L = 10 kΩ		10	40		10	40		10	40	ppm
G = 1–100	R_L = 2 kΩ		10	95		10	95		10	95	ppm
Gain vs. Temperature											
	G =1			10			10			10	ppm/°C
	Gain >1[2]			–50			–50			–50	ppm/°C
VOLTAGE OFFSET	(Total RTI Error = V_{OSI} + V_{OSO}/G)										
Input Offset, V_{OSI}	V_S = ±5 V to ±15 V		30	125		15	50		30	125	μV
Over Temperature	V_S = ±5 V to ±15 V			185			85			225	μV
Average TC	V_S = ±5 V to ±15 V		0.3	1.0		0.1	0.6		0.3	1.0	μV/°C
Output Offset, V_{OSO}	V_S = ±15 V		400	1000		200	500		400	1000	μV
	V_S = ±5 V			1500			750			1500	μV
Over Temperature	V_S = ±5 V to ±15 V			2000			1000			2000	μV
Average TC	V_S = ±5 V to ±15 V		5.0	15		2.5	7.0		5.0	15	μV/°C
Offset Referred to the Input vs.											
Supply (PSR)	V_S = ±2.3 V to ±18 V										
G = 1		80	100		80	100		80	100		dB
G = 10		95	120		100	120		95	120		dB
G = 100		110	140		120	140		110	140		dB
G = 1000		110	140		120	140		110	140		dB
INPUT CURRENT											
Input Bias Current			0.5	2.0		0.5	1.0		0.5	2	nA
Over Temperature				2.5			1.5			4	nA
Average TC			3.0			3.0			8.0		pA/°C
Input Offset Current			0.3	1.0		0.3	0.5		0.3	1.0	nA
Over Temperature				1.5			0.75			2.0	nA
Average TC			1.5			1.5			8.0		pA/°C
INPUT											
Input Impedance											
Differential			10‖2			10‖2			10‖2		GΩ‖pF
Common-Mode			10‖2			10‖2			10‖2		GΩ‖pF
Input Voltage Range[3]	V_S = ±2.3 V to ±5 V	$-V_S + 1.9$		$+V_S - 1.2$	$-V_S + 1.9$		$+V_S - 1.2$	$-V_S + 1.9$		$+V_S - 1.2$	V
Over Temperature		$-V_S + 2.1$		$+V_S - 1.3$	$-V_S + 2.1$		$+V_S - 1.3$	$-V_S + 2.1$		$+V_S - 1.3$	V
	V_S = ±5 V to ±18 V	$-V_S + 1.9$		$+V_S - 1.4$	$-V_S + 1.9$		$+V_S - 1.4$	$-V_S + 1.9$		$+V_S - 1.4$	V
Over Temperature		$-V_S + 2.1$		$+V_S - 1.4$	$-V_S + 2.1$		$+V_S - 1.4$	$-V_S + 2.3$		$+V_S - 1.4$	V
Common-Mode Rejection Ratio DC to 60 Hz with 1 kΩ Source Imbalance	V_{CM} = 0 V to ±10 V										
G = 1		73	90		80	90		73	90		dB
G = 10		93	110		100	110		93	110		dB
G = 100		110	130		120	130		110	130		dB
G = 1000		110	130		120	130		110	130		dB
OUTPUT											
Output Swing	R_L = 10 kΩ,										
	V_S = ±2.3 V to ±5 V	$-V_S + 1.1$		$+V_S - 1.2$	$-V_S + 1.1$		$+V_S - 1.2$	$-V_S + 1.1$		$+V_S - 1.2$	V
Over Temperature		$-V_S + 1.4$		$+V_S - 1.3$	$-V_S + 1.4$		$+V_S - 1.3$	$-V_S + 1.6$		$+V_S - 1.3$	V
	V_S = ±5 V to ±18 V	$-V_S + 1.2$		$+V_S - 1.4$	$-V_S + 1.2$		$+V_S - 1.4$	$-V_S + 1.2$		$+V_S - 1.4$	V
Over Temperature		$-V_S + 1.6$		$+V_S - 1.5$	$-V_S + 1.6$		$+V_S - 1.5$	$-V_S + 2.3$		$+V_S - 1.5$	V
Short Current Circuit			±18			±18			±18		mA

TPS7101 electrical characteristics at I_O = 10 mA, V_I = 3.5 V, $\overline{EN}$ = 0 V, C_O = 4.7 µF/CSR† = 1 Ω, FB shorted to OUT at device leads (unless otherwise noted)

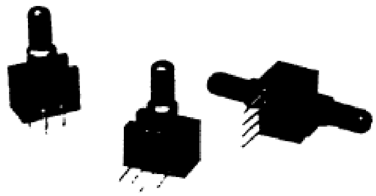
PARAMETER	TEST CONDITIONS‡		T_J	TPS7101Q MIN	TPS7101Q TYP	TPS7101Q MAX	UNIT
Reference voltage (measured at FB with OUT connected to FB)	V_I = 3.5 V,	I_O = 10 mA	25°C		1.178		V
	2.5 V ≤ V_I ≤ 10 V, See Note 1	5 mA ≤ I_O ≤ 500 mA,	−40°C to 125°C	1.143		1.213	V
Reference voltage temperature coefficient			−40°C to 125°C		61	75	ppm/°C
Pass-element series resistance (see Note 2)	V_I = 2.4 V,	50 µA ≤ I_O ≤ 150 mA	25°C		0.7	1	Ω
			−40°C to 125°C			1	
	V_I = 2.4 V,	150 mA ≤ I_O ≤ 500 mA	25°C		0.83	1.3	
			−40°C to 125°C			1.3	
	V_I = 2.9 V,	50 µA ≤ I_O ≤ 500 mA	25°C		0.52	0.85	
			−40°C to 125°C			0.85	
	V_I = 3.9 V,	50 µA ≤ I_O ≤ 500 mA	25°C		0.32		
	V_I = 5.9 V,	50 µA ≤ I_O ≤ 500 mA	25°C		0.23		
Input regulation	V_I = 2.5 V to 10 V, See Note 1	50 µA ≤ I_O ≤ 500 mA,	25°C			18	mV
			−40°C to 125°C			25	
Output regulation	I_O = 5 mA to 500 mA, See Note 1	2.5 V ≤ V_I ≤ 10 V,	25°C			14	mV
			−40°C to 125°C			25	
	I_O = 50 µA to 500 mA, See Note 1	2.5 V ≤ V_I ≤ 10 V,	25°C			22	mV
			−40°C to 125°C			54	
Ripple rejection	f = 120 Hz	I_O = 50 µA	25°C	48	59		dB
			−40°C to 125°C	44			
		I_O = 500 mA, See Note 1	25°C	45	54		
			−40°C to 125°C	44			
Output noise-spectral density	f = 120 Hz		25°C		2		µV/√Hz
Output noise voltage	10 Hz ≤ f ≤ 100 kHz, CSR† = 1 Ω	C_O = 4.7 µF	25°C		95		µVrms
		C_O = 10 µF	25°C		89		
		C_O = 100 µF	25°C		74		
PG trip-threshold voltage§	V_{FB} voltage decreasing from above V_{PG}		−40°C to 125°C	1.101		1.145	V
PG hysteresis voltage§	Measured at V_{FB}		25°C		12		mV
PG output low voltage§	I_{PG} = 400 µA,	V_I = 2.13 V	25°C		0.1	0.4	V
			−40°C to 125°C			0.4	
FB input current			25°C	−10	0.1	10	nA
			−40°C to 125°C	−20		20	

Pressure Sensors

26PC Series

Gage and Differential/Unamplified-Compensated

Temperature Compensated Sensors



FEATURES

- Lowest priced sensor with temperature compensation and calibration
- Variety of gage pressure port configurations - easily and quickly modified for your special needs
- Choice of termination for gage sensors
- Calibrated Null and Span
- Temperature compensated for Span over 0 to 50°C
- Provides interchangeability
- Can be used to measure with vacuum or positive pressure

26PC SERIES PERFORMANCE CHARACTERISTICS at 10.0 ±0.01 VDC Excitation, 25°C

	Min.	Typ.	Max.	Units
Excitation	---	10	16	VDC
Null Offset	−1.5	0	+1.5	mV
Null Shift, 25° to 0°, 25° to 50°C	---	---	±1.0	mV
Linearity, P2 > P1, BFSL	---	±0.25	---	%Span
Sensitivity Shift, 25° to 0°, 25° to 50°C	---	---	±1.0	%Span
1 psi Sensitivity Shift	---	---	±2.0	%Span
Repeatability & Hysteresis	---	±0.20	---	%Span
Response Time	---	---	1.0	msec
Input Resistance	---	7.5 K	---	ohms
Output Resistance	---	2.5 K	---	ohms
Stability over One Year	---	±0.5	---	%Span
Weight	---	2	---	grams

Total error calculation, see page 66

ENVIRONMENTAL SPECIFICATIONS

Operating Temperature	−40° to 85°C (−40° to +185°F)
Storage Temperature	−55° to +100°C (−67° to +212°F)
Compensated Temperature	0° to +50°C (32° to +122°F)
Shock	Qualification tested to 150 g
Vibration	MIL-STD-202. Method 213 (150g halfsine, 11 msec)
Media (P1 & P2)	Limited only to those media which will not attack polyetherimide, silicon and fluorosilicone seal

26PC SERIES ORDER GUIDE

Catalog Listing	Pressure Range psi	Span, mV			Sensitivity mV/psi Typ.	Overpressure psi Max.	Linearity, %Span P2 > P1 Max.
		Min.	Typ.	Max.			
26PCA Type	1	14.7	16.7	18.7	16.7	20	±1.0
26PCB Type	5.0	47	50	53	10	20	±1.0
26PCC Type	15	97	100	103	6.67	45	±1.0
26PCD Type	30	97	100	103	3.33	60	±1.0

DRV102

SPECIFICATIONS

At T_C = +25°C, V_S = +24V, load = series diode MUR415 and 100Ω, and 4.99kΩ $\overline{Flag}$ pull-up to +5V, unless otherwise noted.

PARAMETER	CONDITIONS	DRV102T, F MIN	TYP	MAX	UNITS
OUTPUT					
Output Saturation Voltage, Source	I_O = 1A		+1.7	+2.2	V
	I_O = 0.1A		+1.3	+1.7	V
Current Limit		2	2.7	3.4	A
Under-Scale Current			16		mA
Leakage Current	Output Transistor Off, V_S = +60V, V_O = 0V		±0.01	±2	mA
DIGITAL CONTROL INPUT[1]					
V_{CTR} Low (output disabled)		0		+1.2	V
V_{CTR} High (output enabled)		+2.2		V_S	V
I_{CTR} Low (output disabled)	V_{CTR} = 0V		–80[2]		µA
I_{CTR} High (output enabled)	V_{CTR} = +5V		20[2]		µA
Propagation Delay: On-to-Off			0.9		µs
Off-to-On			1.8		µs
DELAY TO PWM[3]	dc to PWM Mode				
Delay Equation[4]			Delay to PWM ≈ $C_D \cdot 10^6$ (C_D in F)		s
Delay Time	C_D = 0.1µF	80	97	110	ms
Minimum Delay Time[5]	C_D = 0		15		µs
DUTY CYCLE ADJUST					
Duty Cycle Range			10 to 90		%
Duty Cycle Accuracy	49% Duty Cycle, R_{PWM} = 25.5kΩ		±1	±7	%
vs Supply Voltage	49% Duty Cycle, V_S = 8V to 60V		±1	±5	%
Nonlinearity[6]	20% to 80% Duty Cycle		±2		% FSR
DYNAMIC RESPONSE					
Output Voltage Rise Time	V_O = 10% to 90% of V_S		0.25	2.5	µs
Output Voltage Fall Time	V_O = 90% to 10% of V_S		0.25	2.5	µs
Oscillator Frequency		19	24	29	kHz
$\overline{FLAG}$					
Normal Operation	20kΩ Pull-Up to +5V, I_O < 1.5A	+4	+4.9		V
Fault[7]	Sinking 1mA		+0.2	+0.4	V
Sink Current	$V_{\overline{FLAG}}$ = 0.4V		2		mA
Under-Current Flag: Set			5.2		µs
Reset			11		µs
Over-Current Flag: Set			5.2		µs
Reset			11.5		µs
THERMAL SHUTDOWN					
Junction Temperature					
Shutdown			+165		°C
Reset from Shutdown			+150		°C
POWER SUPPLY					
Specified Operating Voltage			+24		V
Operating Voltage Range		+8		+60	V
Quiescent Current	I_O = 0		6.5	9	mA
TEMPERATURE RANGE					
Specified Range		–55		+125	°C
Storage Range		–55		+125	°C
Thermal Resistance, θ_{JC}					
7-Lead DDPAK, 7-Lead TO-220			3		°C/W
Thermal Resistance, θ_{JA}					
7-Lead DDPAK, 7-Lead TO-220	No Heat Sink		65		°C/W

G HU Z 040 M20 D03

Performance and dimensions for type G HU Z 040 M20 D03

G HU Z 040 M20 D03		
Duty rating (ED)	(%)	100
Stroke (s)	(mm)	10
Work rating (A_N)	(Ncm)	10.8
Power consumption (P_{20})	(W)	12.9
Ambient temperature (δ_{11})	(^{0}C)	35
Frequency of operating (S_h max)	(1/h)	25000
Closing time (t_1)	(ms)	85
Opening time (t_2)	(ms)	50
Armature weight (m_A)	(kg)	0.078
Solenoid weight (m_M)	(kg)	0.46
Radial bolt load (max) allowable (approx) static	(N)	1500
stroke	(N)	8

LINEAR ACTUATORS: AIR CYLINDERS
SERIES C85

TECHNICAL SPECIFICATIONS

<table>
<tr><td colspan="2">Bore size (mm)</td><td>8</td><td>10</td><td>12</td><td>16</td><td>20</td><td>25</td></tr>
<tr><td colspan="2">Piston rod dia. (mm)</td><td>4</td><td>4</td><td>6</td><td>6</td><td>8</td><td>10</td></tr>
<tr><td colspan="2">Piston rod thread</td><td>M4X0.7</td><td>M4X0.7</td><td>M6X1</td><td>M6X1</td><td>M8X1.25</td><td>M10X1.25</td></tr>
<tr><td colspan="2">Ports</td><td>M5</td><td>M5</td><td>M5</td><td>M5</td><td>G⅛</td><td>G⅛</td></tr>
<tr><td colspan="2">Action</td><td colspan="6">Double acting: single + double rod, non rotating
Single acting: sring return + extended non-rotating</td></tr>
<tr><td colspan="2">Fluid</td><td colspan="6">Air</td></tr>
<tr><td colspan="2">Proof pressure</td><td colspan="6">15 Bar</td></tr>
<tr><td colspan="2">Max. operating pressure</td><td colspan="6">10 Bar</td></tr>
<tr><td colspan="2">Min. operating pressure Double acting</td><td>1 Bar</td><td colspan="2">0.8 Bar</td><td colspan="3">0.5 Bar</td></tr>
<tr><td colspan="2">Min. operating pressure Single acting</td><td>2.2 Bar</td><td colspan="2">1.8 Bar</td><td>1.3 Bar</td><td colspan="2">2.3 Bar</td></tr>
<tr><td colspan="2">Ambient and fluid temperature</td><td colspan="6">-20°-+80°C (Built-in magnet type: Max. 60°C)</td></tr>
<tr><td colspan="2">Cushion</td><td colspan="6">Rubber cushion (standard)/Air cushion</td></tr>
<tr><td colspan="2">Lube</td><td colspan="6">None (Non-lubricated)</td></tr>
<tr><td colspan="2">Piston speed</td><td colspan="6">50-750mm/S</td></tr>
<tr><td colspan="2">Allowable kinetic energy (Nm)</td><td>0.02</td><td>0.03</td><td>0.04</td><td>0.09</td><td>0.27</td><td>0.40</td></tr>
<tr><td rowspan="3">Construction</td><td>Rod cover</td><td colspan="6">Alluminium alloy (White alumite)</td></tr>
<tr><td>Cylinder tube</td><td colspan="6">Stainless steel</td></tr>
<tr><td>Piston rod</td><td colspan="5">Stainless steel</td><td>Carbon steel
Hard chrome finish</td></tr>
</table>

Digital linear actuator

Specification

Standard models		**L92121-P2** **K92121-P2**	**L92111-P1**	**L92211-P2** **K92211-P2**	**L92411-P2**
Maximum linear force	N	7.23	12.5	20.9	88
Min. holding force (de-energised)	N	11.13	16.6	11.13	88
Linear travel per step	ins./mm	0.002 / 0.0508	0.001 / 0.0254	0.001 / 0.0254	0.001 / 0.0254
Typical backlash	Steps	2	2	2	2
Maximum linear travel: **L92000 series** using standard screw using extended screw **K92000 series**	 mm mm mm	 47.6 259 12.7	 47.6 259 N / A	 47.6 215 22.2	 76.2 233 N / A
Maximum Pull-in rate	Steps/sec.	380	425	425	275
Maximum Pull-out rate	Steps/sec.	650 *	700 *	700 *	400 *
Bearing construction		Radial Ball	Radial Ball	Radial Ball	Radial Ball
Mass	Kg	0.0425	0.0425	0.198	0.45
Nominal Voltage (L/R Drive)	Vdc	12	5	12	12
Resistance per phase	Ohms	84	15	58	25
Current per phase	Amps	0.146	0.333	0.208	0.453
Inductance per phase	mH	29	5.0	30	25
Suitable drives		SAA 1027 MSE422 EM162	MSE422 EM162 TM162C	SAA 1027 MSE422 EM162	MSE422 MSE542 EM162 TM162C